SELF POWERED WRIST EXTENSION ORTHOSIS

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M. K. Singer

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Abstract

One of the most devastating effects of tetraplegia is the inability to grasp and manipulate everyday objects necessary to living an independent life. Currently surgery is widely accepted as the solution to improve hand functionality. However, surgery becomes difficult when the user has paralysed wrists as is the case with C5 tetraplegia. The aim of this research was to develop a solution which provided controlled wrist flexion and extension which, when combined with surgery, achieves a 'key pinch' grip. This particular grip is critically important for people with C5 tetraplegia as it is used for countless grasping activities, necessary on a day-to-day basis.

A systematic design process was used to evolve the solution to provide controlled wrist flexion and extension. Concept brainstorming identified four alternative solutions which were evaluated to find the preferred concept. The chosen solution was called the *Self Powered Wrist Extension Orthosis,* more commonly referred to as the 'orthosis'. This concept contained a *shoulder harness* which provided both energy and control to the *wrist harness,* which in turn changed the wrist position. The orthosis was developed with the use of a mathematical model which theoretically predicted the functional performance by comparing the required force needed to move the wrist harness to the achievable force supplied by the user's shoulders. Using these parameters, the orthosis was optimized using the matlab Nelder-Mead algorithm which adjusted the wrist harness geometries to maximize the functional performance.

A prototype was constructed and tested with the help of two participants who when combined, achieved an average of 18.5° of wrist rotation. The theoretical model however predicted an average range of motion of 28.4°. The discrepancy found between the theoretical and experimental result can be contributed to incorrect assumptions in the theoretical model. This included unaccounted friction and inaccurate modeling of the orthosis dynamics. The feedback from potential users of the orthosis was enthusiastic and encouraging especially towards the simplicity, usability and practicality of the design.

Chapter 1 Introduction

1.1 Introduction

Spinal cord injury (SCI) is one of the most complicated injuries the human body can sustain. However, unlike most injuries, damage to the spine does not naturally repair and the person with the injury must prepare themselves for a dramatic change in lifestyle. Tetraplegia is a type SCI located in the neck region of the spine and results in paralysis of the upper body. Due to the decrease in functionality of the upper extremities such as the arms and hands, it becomes difficult to perform simple tasks such as writing with a pen or opening a door. These everyday activities which are necessary to function on a day-to-day basis are commonly referred to as ‘activities of daily living’.

C5 tetraplegia occurs as a result of damage around the 5^{th} cervical vertebrae. The result of the injury includes loss of leg; abdomen; hand; and wrist function. A person with C5 tetraplegia has the ability to move their arms freely, but due to the lack of hand function, picking up or manipulating objects is very difficult. To resolve this, most solutions aim to restore a key grip which is produced when the thumb and the index finger pinch together. One such solution is a surgical operation which pulls the thumb to the side of the index finger as the user extends their wrist, and this is known as a passive key pinch secondary to wrist extension. However this operation is only applicable to those with C6 tetraplegia because they have wrist control, however a person with C5 tetraplegia does not have wrist control and therefore the operation is not performed. The aim of this research was to develop a solution which would enable a person with C5 tetraplegia with the means to control their wrist movement, and with the added surgery, achieve a key pinch grip. Sections 1.2 through to 1.6 provides background information into medical terminology; anatomy; and physiology which is required to understand this research.

Mission Statement:

“provide controlled wrist extension and flexion for a person with C5 tetraplegia”

1.2 Medical Terminology

Anatomy and physiology provide the foundation for understanding the body and its functions. Anatomy is defined as the science of body structures while physiology is the science of body functions. Anatomy requires a common language to accurately explain positions on the human body and is established by describing the body in the 'anatomical position'. The definition of the anatomical position is looking at another human from the anterior (front) view, the head is facing forwards, the person is standing straight up, the hands are to the side of the body with the palms facing forwards, the legs are straight and the toes are pointing forward. The second view is the posterior view which is viewing the body from behind. Figure 1-1 shows the directional terms of the human body from the anterior view. These terms are used to precisely locate various parts of the body relative to one another.

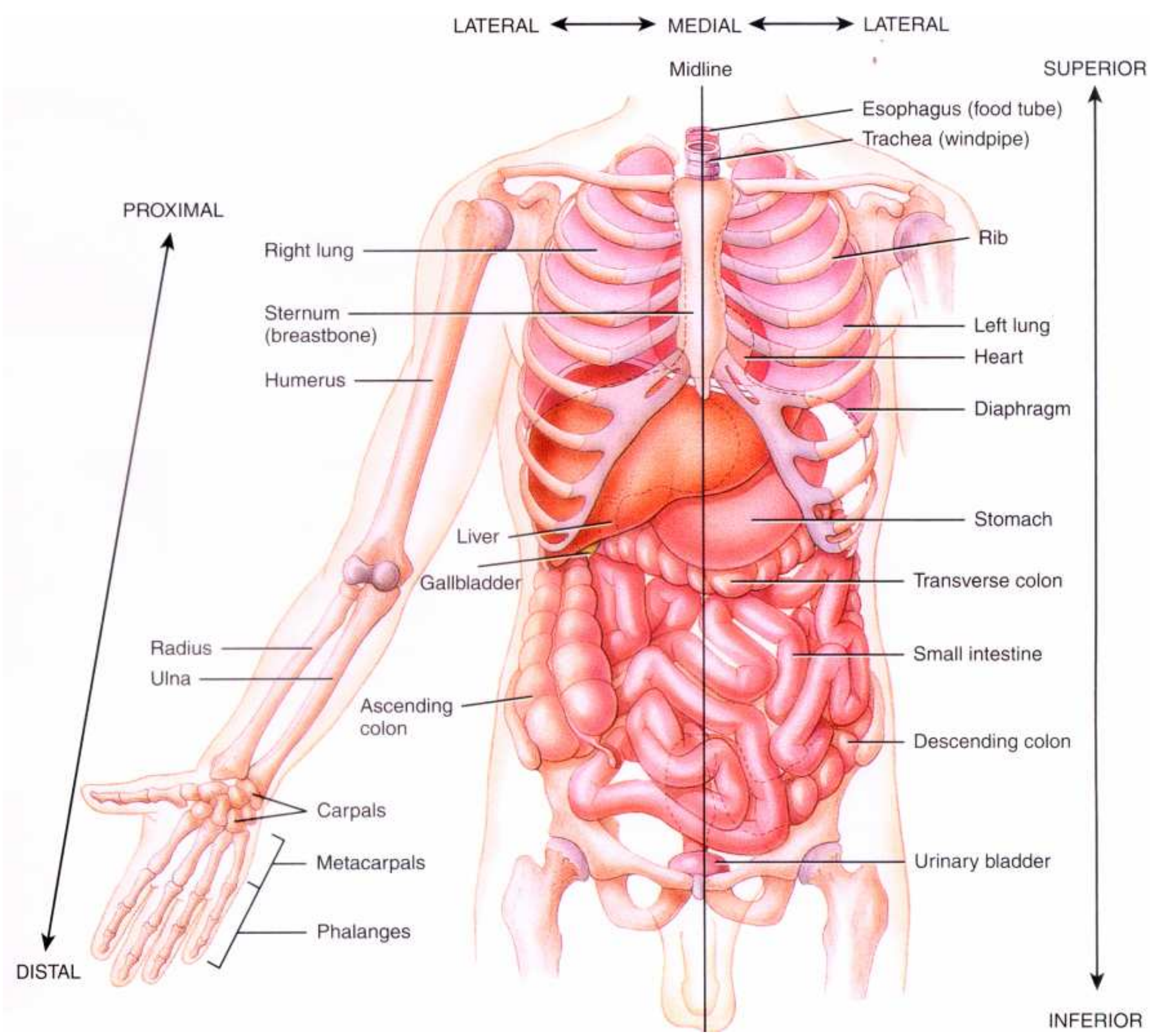


Figure 1-1 Directional terms of the human body (Tortora and Grabowski, 2003)

1.3 What is Spinal Cord Injury

Spinal cord injury (SCI) is damage to the spinal cord that results in a loss of function like mobility or feeling. Frequent causes of damage are trauma (car accident, gunshot, falls, etc.) or disease (polio, spina bifida, MS). The spinal cord does not have to be severed in order for a loss of functioning to occur. In fact, for most people with SCI, the spinal cord is intact, but the damage to it results in a loss of functioning.

Tetraplegia, also called quadriplegia, is a SCI located at the cervical (neck) region of the spine and results in paralysis of all four limbs. The cervical region of the spine contains the nerve roots which extend into the arms and hands. Therefore depending on where the spine is injured in the cervical region, determines the extent of paralysis in the arms and hands. An injury which is located high in the cervical region will result in almost complete paralysis of the arms and hands while a lower level injury in the cervical region may result in paralysis of only the fingers. It is important to note that all the nerve roots below the injury will also be damaged and hence a person with tetraplegia will inherently have full paraplegia as well[1] i.e. paralysis of the legs.

[1] This is only applicable to someone with ‘complete’ SCI.

1.4 The Spinal Cord

The Spinal Column starts at the neck and ends at the tailbone. It consists of 33 individual bones called vertebrae. Each vertebra is held together by disks, ligaments and muscles. The spine provides support to the upper body such as the head and torso. It also protects the internal organs, and accommodates the spinal cord which connects the brain with every part of the body. The spine has the ability to bend, rotate, and slip, with only a thin layer of skin protecting it from a physical force and can therefore be easily be damaged, jeopardizing the spinal cord within.

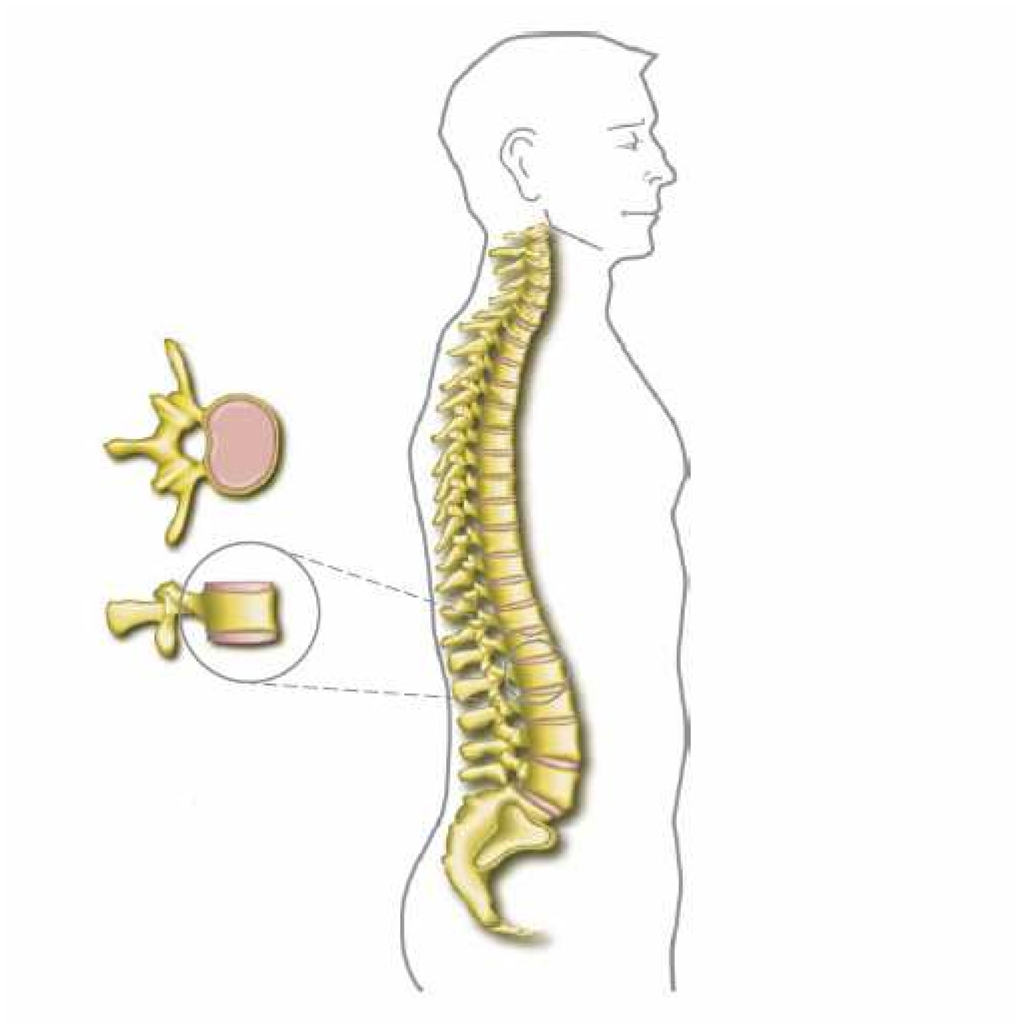


Figure 1-2 Spinal Cord with detailed vertebrae (Verkaaik, 2002)

The spine can be separated into four sections. The cervical region (cervic = the neck) containing the nerve pairs controlling sections of the internal organs like the diaphragm and upper extremities. Further down the spine, is the thoracic (thorax = chest) region that contains the nerve pairs which extend into the abdomen and chest muscles providing core balance. The lumbar (lumb = loin) section provides the nerve pairs which extend into the lower extremities, and lastly the sacrum (SA-krum = sacred bone) consists of five fused sacral vertebrae. The nerve pairs extending out of the sacrum control the anal and bladder sphincter.

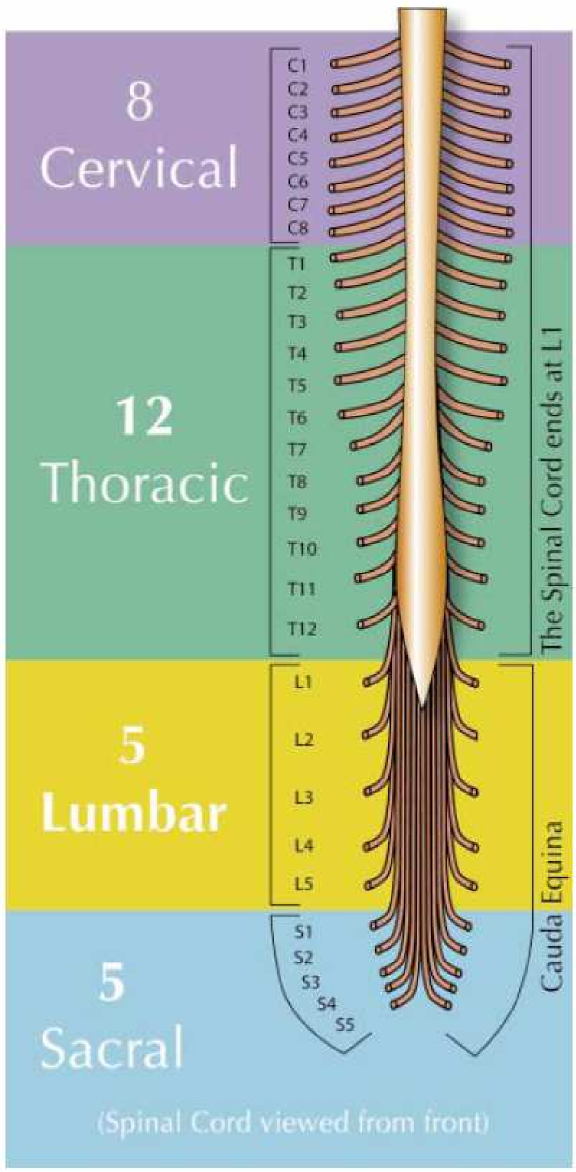


Figure 1-3 Spinal Column (Verkaaik, 2002)

SCI occurs when damage to the spinal cord blocks information from the brain to the body and vice-versa. There are several forms of SCI such as a break, tear, rip or crush that is generally caused by a physical force which is known as a traumatic injury. Figure 1-4 below shows various examples of impairment to the spine. In some cases the spine is completely severed and results in complete paralysis of all the limbs below the level of injury. However the more common injury is incomplete SCI and occurs when only a section of the spine is impaired. The level of functionality for this type of SCI is dependant on how the spinal cord was impaired. Other forms of injury can be caused by diseases, such as multiple sclerosis, polio, or a malignant growth. It is important to note that a person can "break their back or neck" yet not sustain a SCI if only the bones around the spinal cord (the vertebrae) are damaged, but the spinal cord is not affected. In these situations, the individual may not experience paralysis after the bones are stabilized.

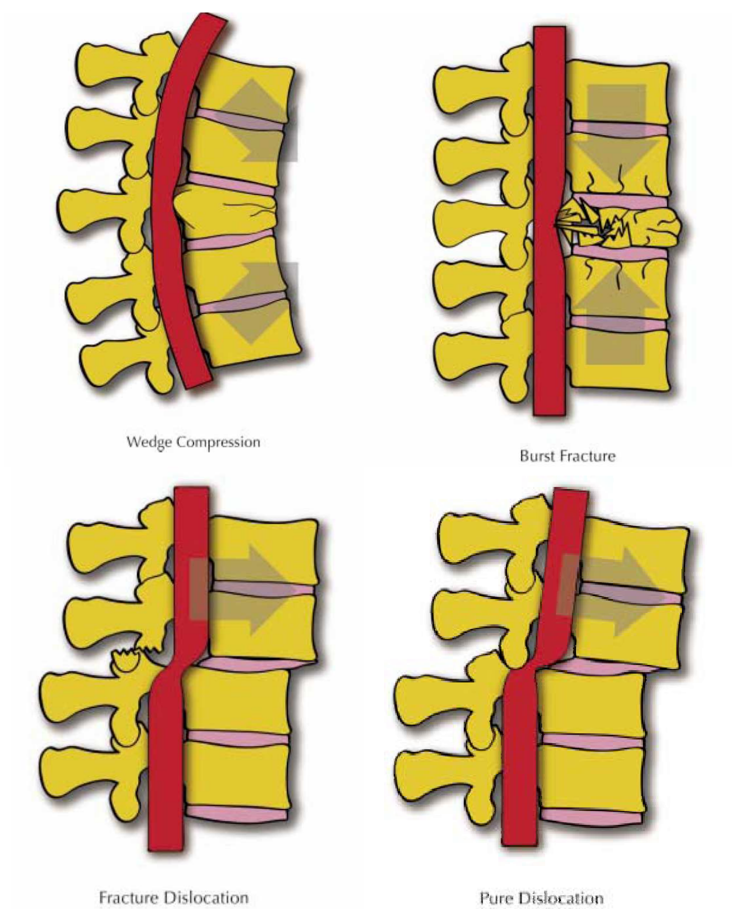


Figure 1-4 Different causes of impairment to the spine (Verkaaik, 2002)

1.5 Muscles of the Hand and Forearm

Figure 1-5 shows the extensor muscles of the hand and forearm. The left side of the figure shows the superficial muscles while the right side shows the deep muscles. The key muscles to note are the brachioradialis, extensor carpi radialis brevis, and flexor pollicis longus (shown in Figure 1-6). These muscles are commonly used in surgical operations to restore hand functionality and are referred to later in the report.

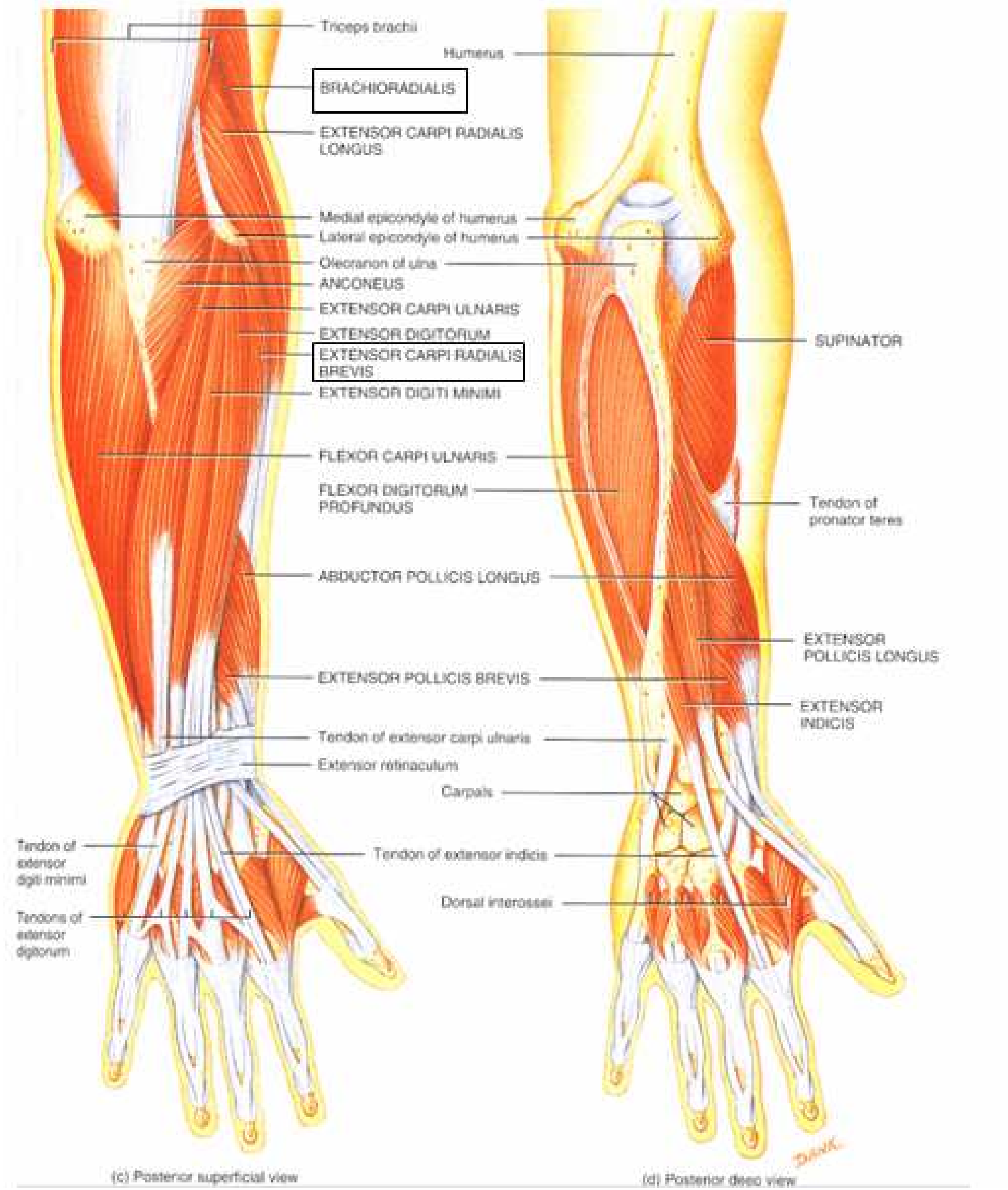


Figure 1-5 Extensor muscles which move the hand and fingers (Tortora and Grabowski, 2003)

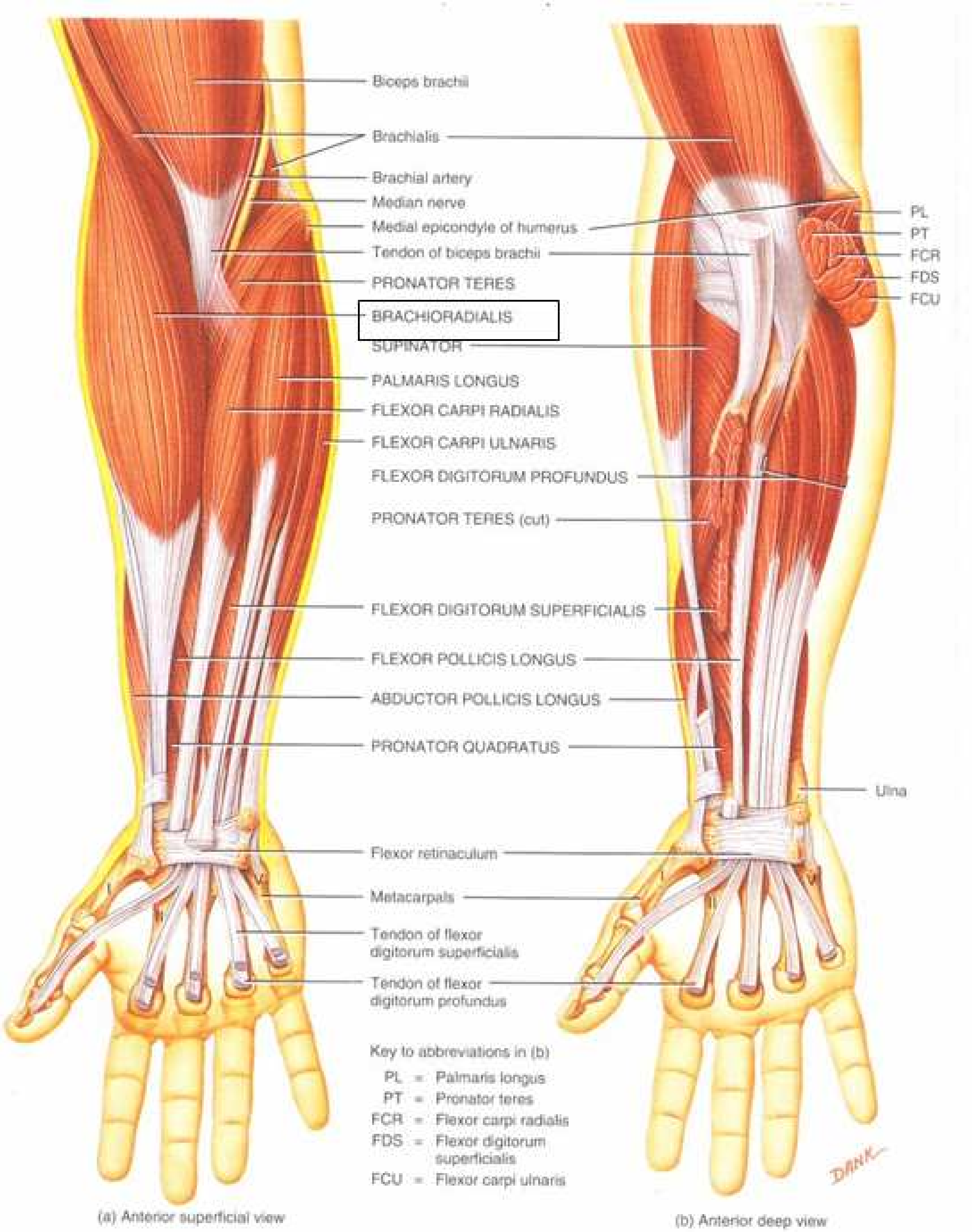


Figure 1-6 Flexor muscles which move the hand and fingers (Tortora and Grabowski, 2003)

1.6 Major motions of the Upper Extremities

The upper extremities contain the shoulder, arm, forearm and hands. The Range of Motions (ROM) critical for the upper extremities are explained in the following diagrams. The values shown are the average for an able bodied person and are taken from the Humanscale data sheets (Diffrient et al., 1981).

Key movements:

Flexion: decrease in angle between the articulating bones

Extension: decrease in angle between the articulating bones

Abduction: Movement of a bone away from the midline

Adduction: Movement of a bone toward the midline

Elevation: Superior movement of a body part

Depression: inferior movement of a body part

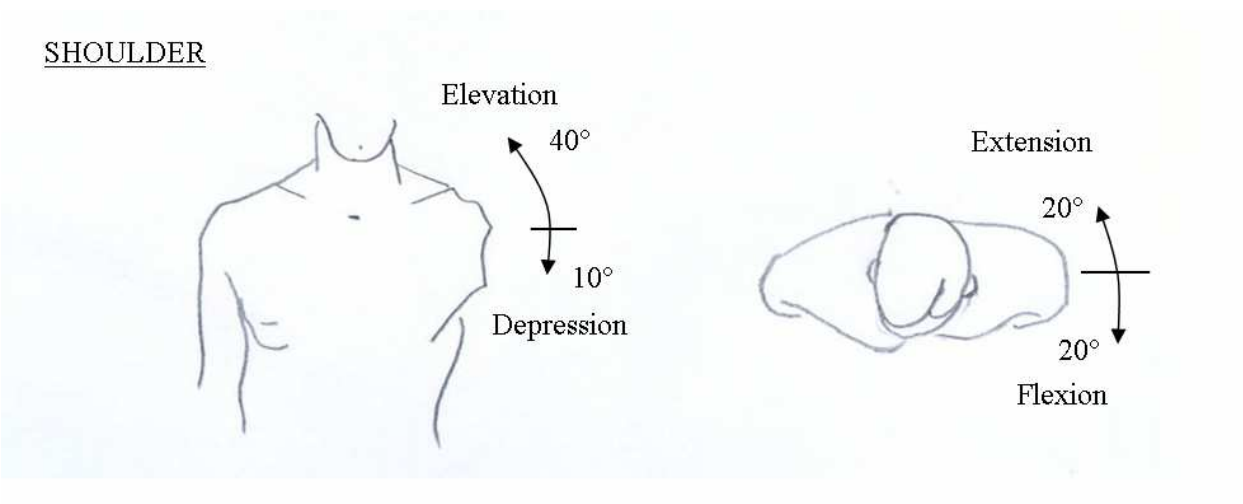


Figure 1-7 motions of the shoulder joint

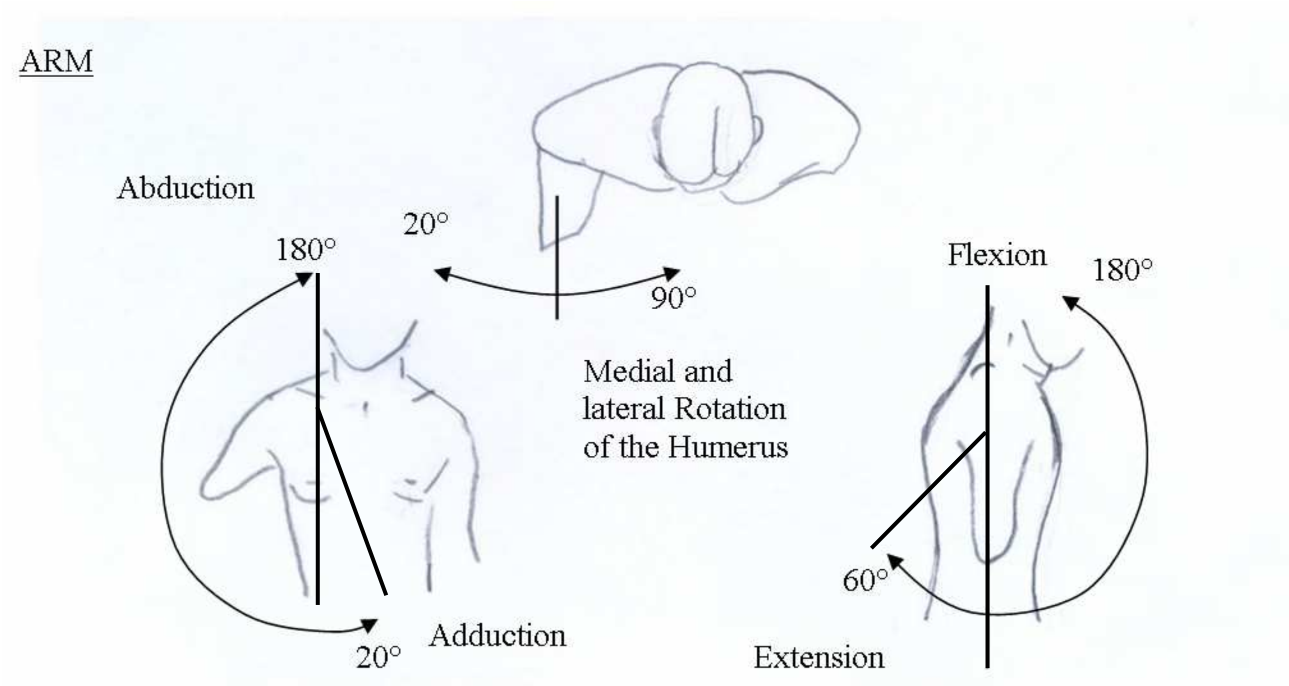


Figure 1-8 motions of the arm

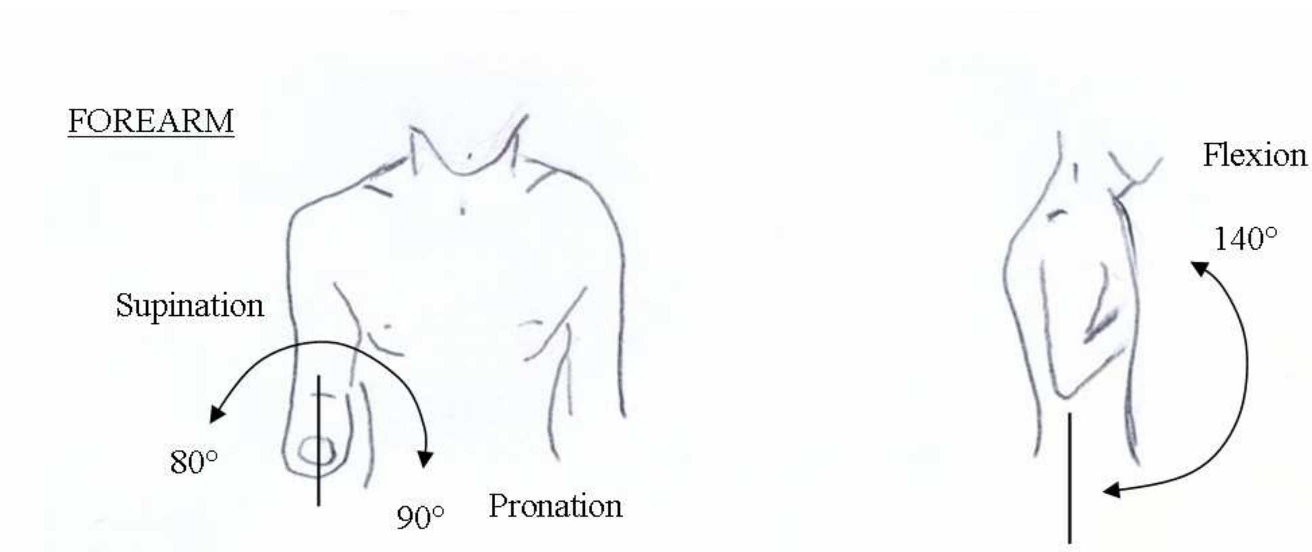


Figure 1-9 motions of the forearm

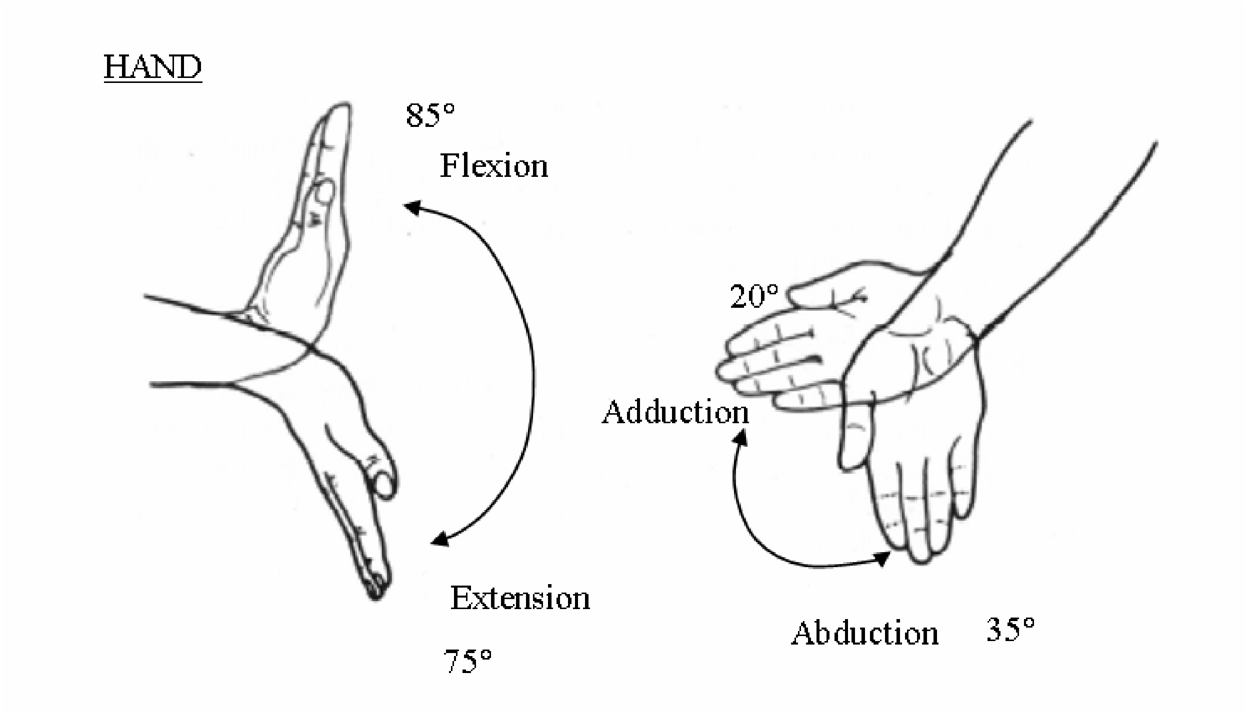


Figure 1-10 motions of the hand

Chapter 2 Literature Review

The purpose for the literature review was to demonstrate the need for a solution to provide a person with C5 tetraplegia with the means to grasp an object. The literature review analyzes the most common level of SCI and the difficulties of living with C5 tetraplegia. This is followed by an analysis of the current solutions available which help people with tetraplegia with the ability to grasp objects. The final section addresses why a new solution is necessary and the steps required for its development.

2.1 Spinal Cord Statistics

The American National Spinal Cord Injury Statistical Center (NSCISC) supervises and directs the collection, management and analysis of the world's largest spinal cord injury database. Figure 2-1 below demonstrates the percentages of the various levels of spinal cord injury. From this graph it is clear that C5 tetraplegia is the most common level of spinal cord injury at 14.8% closely followed by C4 tetraplegia with 13.8%.

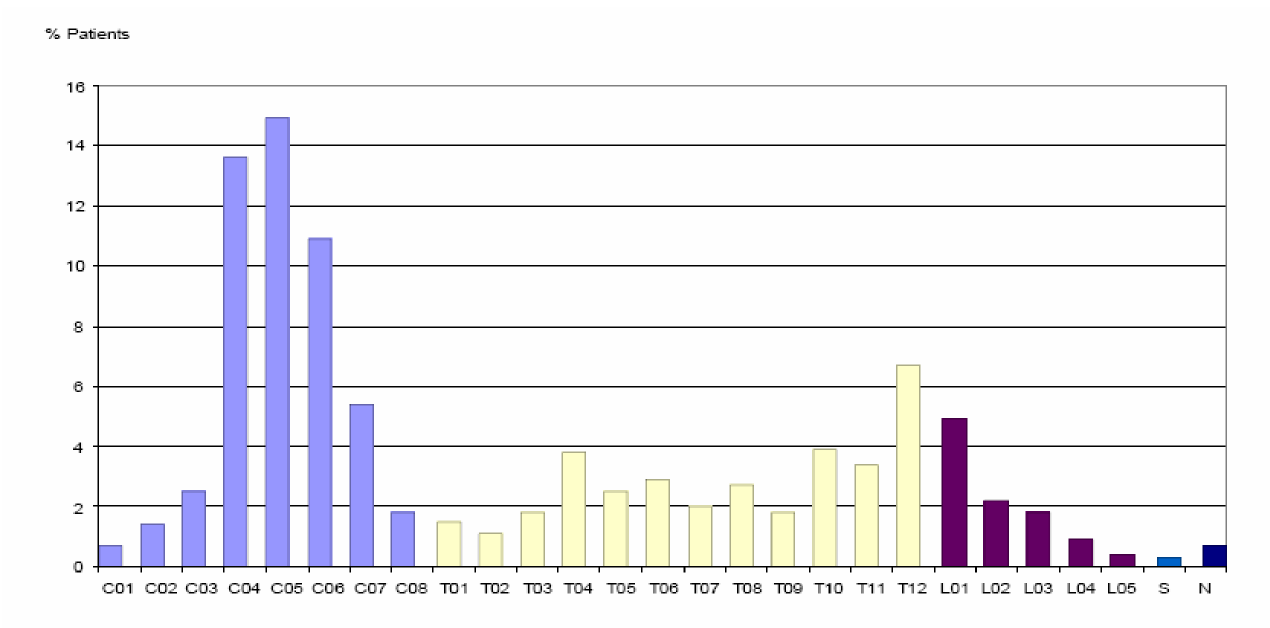


Figure 2-1 Percent of patients by neurological level of legion at discharge (NSCISC, 2004)

Spinal cord injuries primarily affect young male adults with an average age at injury of 28.7, and most injuries occurring between the ages of 16 and 30. The most common cause of injury is vehicle crashes 47.5%; followed by falls 22.9%; violence 13.8%; sports 8.9%; and other unknown causes 6.8%. These statistics are only a representation of the world wide etiology which is unknown.

Figure 2-2 shows the neurological category at discharge in association with type of accident. A feature of interest is the sports injury statistics. Unlike vehicular accidents and falls where accidents results in paraplegia and tetraplegia, sport injuries generally result in only tetraplegia which are caused by injuries from diving, gymnastics, motocross and alpine skiing (Schmitt and Gerner, 2001). It is estimated that the annual incidence of SCI, not including those who die at the scene of the accident, is approximately 40 cases per million.

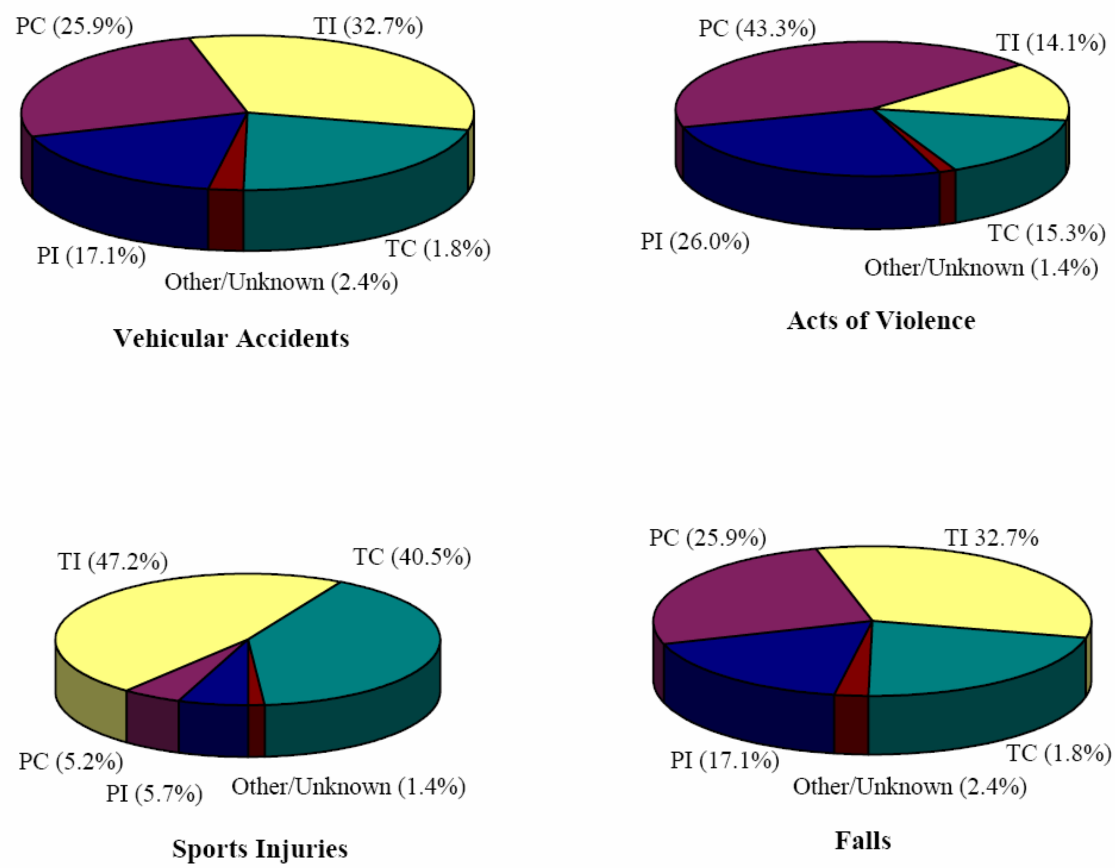


Figure 2-2 Neurological category at discharge by grouped etiology (NSCISC, 2004)
(T-Tetraplegia. P-Paraplegia. I-Incomplete. C-Complete)

There is a high cost associated with caring for someone with a spinal cord injury. Figures suggest that the lifetime cost for a person with C5 tetraplegia who is injured at age 25 exceeds $1.5 million U.S. and these figures do not include any indirect costs such as losses in wages, fringe benefits and productivity (NSCISC, 2004). These high costs are explained in Section 2.2 which details the functionality of a person with C5 tetraplegia, and Section 2.3 shows the current solutions available which are ultimately aimed at reducing these costs.

2.2 C5 Functionality Analysis

The disabilities associated with C5 tetraplegia are complete loss of function to the lower extremities including the abdomen and bowel. The upper extremities however, are only partially affected such as the wrist, hand, fingers and thumb. The shoulders, deltoids and biceps are all functioning but they are limited in strength. Suppination of the hand is possible with the use of the biceps, but pronation requires triceps and therefore can only be achieved by lifting the elbow and allowing the hand to ‘fall’ over using gravity.

A person with C5 tetraplegia is capable of breathing without a ventilator however they still require assistance with coughing; washing; dressing and bowel and bladder management. Complete domestic care is required such as cleaning, washing and preparation of meals. Independent transport is possible with the use of an electric wheelchair, and on very smooth and hard ground it is possible for them to use a manual wheelchair. They provide motion by using their biceps while gripping the wheel with the friction generated between the medial side of their palm and the tire. Because only a small force can be transferred using this technique any obstacle such as a ramp is difficult to overcome.

After rehabilitation, most people with C5 tetraplegia will rejoin the workforce but require an occupation with minimal manual effort. The person will be able to type on a computer with the help of a hand brace and typing stick, or alternatively use voice recognition software. Self-feeding is possible using a feeding strap and fork. It seems the greatest disadvantage for people with C5 tetraplegia is the loss of hand function. According to many researches 75% of persons with tetraplegia would rather regain hand function than any other function that they have lost, including bowel function, bladder function, sexual function, and even the ability to walk

(Moberg, 1975). Simple tasks, such as holding a drink, or writing with a pen, are virtually impossible without the use of the hand. It is for this reason that researches for many years have tried to develop solutions to provide hand function back to those with C5 tetraplegia.

2.3 *Current Solutions*

The objective for most solutions aimed at helping someone with C5 tetraplegia is to provide the person with ability to grasp objects that are experienced on a day-to-day basis. These solutions include; surgery; functional electrical stimulation (FES), and mechanical exoskeletons. The solutions generally provide an opposable digit which enables a person to grip and manipulate objects. The most basic grip is a ‘key pinch’ grip (Figure 2-3) which is achieved by activation of the thumb with opposition from the lateral side of the index finger. The ‘tripod’ grip is another simple grip that is formed by the opposition of the index and middle finger against the thumb. Reports suggest that with the use of a tripod and key grip, as much as 80% of daily tasks can be achieved, and a key grip alone can achieve up to 50% of these tasks.

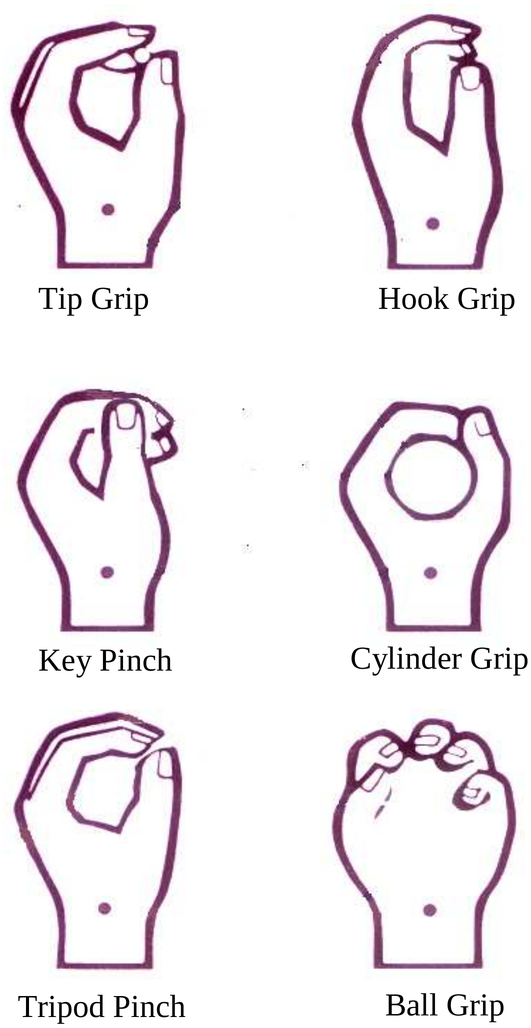


Figure 2-3 Types of hand grips (Diffrient et al., 1981)

2.3.1 Surgical Solutions

The current surgical solution to improve the functionality for a person with C5 tetraplegia is a combination of two surgical procedures. Firstly, wrist extension is required to provide control of the wrist, followed by an operation to the thumb so that when the wrist extends, the thumb forms a key pinch grip. Wrist extension is generally achieved by taking an active muscle and transferring it onto a wrist extensor muscle. But for a person with C5 tetraplegia there are very few active muscles that can be used for this operation. However, one muscle called the Brachioradialis (located in the forearm that primarily acts to flex the forearm, Figure 1-5) can be transferred onto the extensor carpi radialis brevis (a wrist extensor muscle, Figure 1-5) which provides active wrist extension (Schindler L et al 1994). The second operation provides a key pinch grip secondary to wrist extension. This is achieved through a tenodesis[2] of the flexor polices longus which pulls the thumb towards the lateral side of the index finger as the wrist extends. This operation requires a small incision to the anterior side of the wrist where the flexor polices longus tendon is cut and sutured onto the bone.

The success of using surgery as a means to improve functionally, can be attributed to the fact that it requires no external power supply; there are no un-natural features; the solution is permanent; it ‘feels’ normal; and the control and activation is neurologically controlled. It is therefore unsurprising that surgery has been the leading solution for people with tetraplegia. But even though surgical procedures function very well, there are some key setbacks.

A suitable candidate for the operation requires a certain level of functionality, for example, the brachioradialis muscle is necessary for wrist extension. In some cases, a person with C5 tetraplegia may find they have a strong brachioradialis, whereas others will not (Freehafer, 1991). Without this active muscle to transfer there is little reason to perform the operation. Surgery also requires a lot of time post surgery to allow for rehabilitation and rest. For some operations, this can take up to 28 months (Schindler et al., 1993) and understandably some people are not prepared to persevere with hospitilzation, surgery, and rehabilitation (Mohammed et al., 1992).

[2] Tenodesis is the operation of suturing the end of a tendon to a bone

2.3.2 Mechanical Exoskeleton Solution

The term *exoskeleton* when used in the context of assistive devices is generally considered as an externally powered device which couples itself to the body structure. Exoskeletons are continually improving in design and technology, having developed from single to multi-degrees of freedom structures with intelligent design and control. A good example of the current technology is the University of Washington Upper Limb Powered Exoskeleton shown in Figure 2-4. This exoskeleton provides a full range of arm motion and is controlled using potential signals produced by active muscles. This method of control is called *electromyogram* signals, more commonly known as EMG.

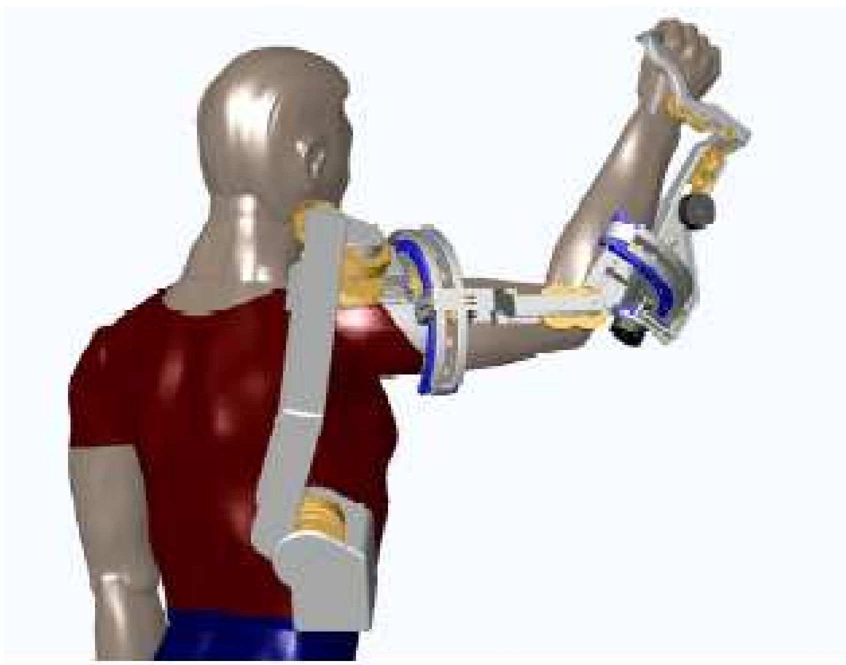


Figure 2-4 University of Washington Upper Limb Powered Exoskeleton (Perry and Rosen, 2006)

The current market focus seems to be towards producing a controllable active exoskeleton. These devices are becoming more sophisticated with improved control, but as yet, they are still not used on a regular basis. The current trend with these devices is to use EMG control with mechanical motion. The Power Grip, shown in Figure 2-5 is a fairly typical concept which utilizes motors and rigid links. Functionally this concept works very well, but practically it is heavy, lacks control, produces noise, and is non discreet.

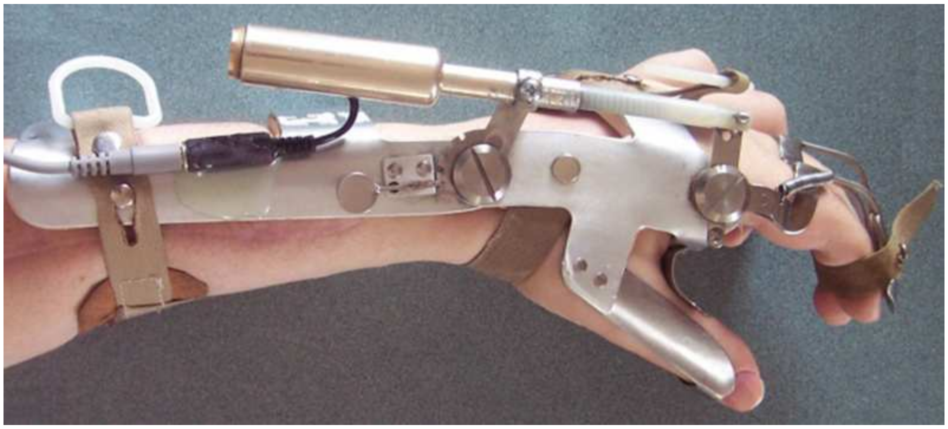


Figure 2-5 Power Grip (Felling, 2006)

2.3.3 Functional Electrical Stimulation

Functional electrical stimulation commonly referred to as FES is a more recent advancement in rehabilitation design. It works by applying a potential to a muscle, causing it to contract and create movement. There are two systems currently available which apply FES to wrist extensor muscles. The Neuromuscular Electrical Stimulation System (NESS) works by fixating the wrist in an extended position. By applying FES to contract certain muscles, a functional grasp is obtained. Though the system appears elegant, there are some key limitations to its functionality like difficulty positioning of electrodes, poor activation and control, low achievable grasping forces, and muscle fatigue. The device is shown in Figure 2-6.

Figure 2-6 the NESS functional electrical stimulation device (Alon and McBride, 2003)

The second option is the NeuroControl Freehand System which is one of the most advanced systems available that is specifically designed for people with C5 tetraplegia. The advantage of this system is that it is almost completely hidden; however it requires surgery to implant electrodes and transmitters under the skin. The device is activated and controlled with the use of a mechanism placed on the auxiliary[3] shoulder. Certain movements of this shoulder relates to certain hand grips that are determined using a controller. The device was commercially produced however it is currently undergoing improvements.

2.3.4 Robotics

Robots offer exciting new possibilities for people with tetraplegia. The functionality of a robot is virtually endless given the current technological trend. The ASIMO humanoid robot created by Honda is capable of running, communicating and carrying loads. These robots could easily be adapted to hold a book, or carry a drink, and perform tasks to help people with tetraplegia.

2.3.5 Stem Cells

For years researches have been trying to use stem cells to repair damage to the spine. Stem cells have the ability to continually reproduce themselves while maintaining the capacity to give rise to other more specialized types of cells. In the case of a damaged spine, stem cells can be used to help repair the cells in the spinal cord. There are several methods that are currently being developed, but due to the complexity of stem cells and the current ethical debate, this research will take some time before it becomes a viable option.

[3] Auxiliary refers to the opposite

2.4 Solution Overview

The literature review demonstrated that C5 tetraplegia was the most common level of SCI primarily affecting young male adults. The cost associated with this injury is high, which is significantly affected by the inability to grasp and release objects which reduces a person's independence. There are several solutions which are currently available to restore hand functionality however evidence suggests that the current solutions are only used by a few select groups. It seems that some people select surgery while others utilize assistive devices, or further still, some people prefer to live with their disabilities and do without any intervention. While one solution is suitable for a certain set of candidates, the same solution is not suitable for another. The current solutions to restore hand functionality seem to be divided throughout the market leaving a potential to design a solution which fulfills this gap.

2.5 What Makes a Successful Solution?

Gathering the requirements necessary to create a successful solution was difficult however was resolved with the help of International Classification of Disability and Health (ICF) which is used as a tool to predict a person's 'quality of life'. The hypothesis was that if the solution could demonstrate a significant improvement in the quality of life, then it would be accepted by the user. The aim was to extract from the ICF the requirements necessary to create an acceptable solution; this is shown in Chapter 3, The Design Requirement Specification.

Chapter 3 Design Requirement Specification

The aim of the design requirement specification was to develop a list of all the parameters necessary to create a successful solution. The information was gathered from physiotherapists, surgeons, engineers, users of the solution, and the International Classification of Disability and Health (ICF).

3.1 The International Classification of Disability and Health (ICF)

In 1980 the World Health Organization (WHO) published a model designed to represent a person's 'Quality of Life'. This model was revised several times and is now referred to as the International Classification of Functioning, Disability and Health (ICF). The scope of the ICF (Figure 3-1) is broken into two parts. Part 1 consists of three factors; Body Functions and Structures, Activities, and Participation. Part 2 covers two Contextual Factors; Environmental, and Personal Factors. These five factors are interdependent but each can be measured individually using certain tools specific for the particular health condition which is being measured. Sections 3.1.1 through to 3.1.5 describe the tools used to measure each of the factors and from these tools the design requirements were extracted for the solution.

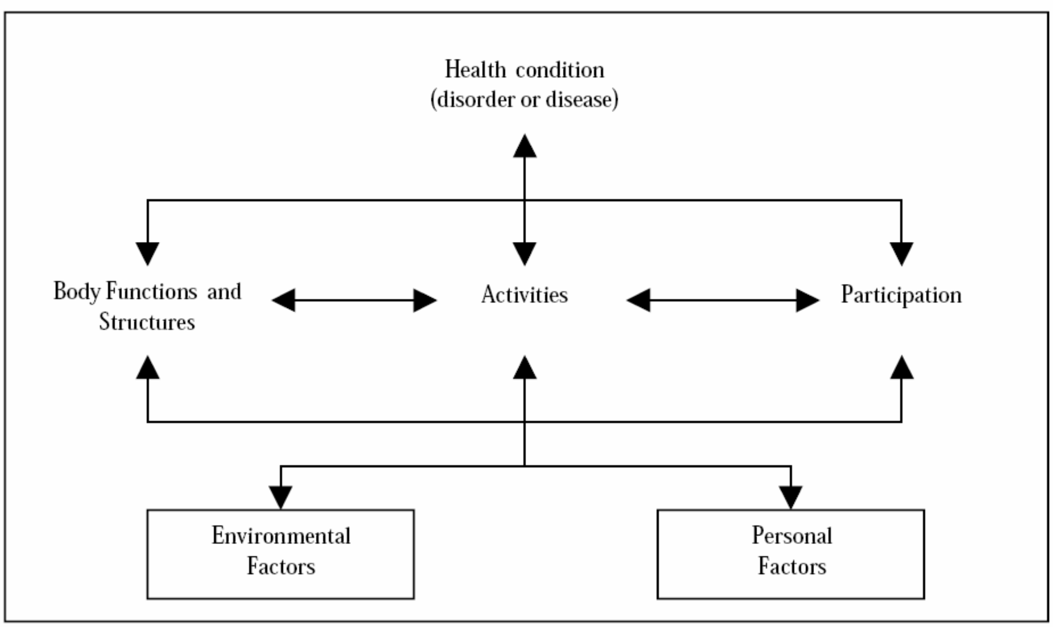


Figure 3-1 Interactions between components of the ICF (WHO, 1002)

3.1.1 Body Functions and Structures

Body Functions and Structures is the measurement of both physiological functions and anatomical parts such as organs, limbs, and their components. The Grasp and Release Test (GRT), shown in Appendix A, P107, was found to be the ideal measurement tool to determine the key performance criteria of the Body Functions and Structures (Sinnot et al., 2004). In the test, users grasped, moved, and released one of six different objects as many times as possible in five 30-second trials for each object. Three of the objects required a key grip (peg, paperweight and fork) and the remaining three required a tripod grip (block, can and videotape). The peg and block represented small, light objects such as pens or finger foods. The paperweight, can, and videotape represented larger, medium weight objects such as a glass or a book. The fork used a spring-loaded piston and required 4.4N of vertical force to depress it sufficiently to represent stabbing pieces of fruit, meet and vegetables (Wuolle and Doren, 1994).

These findings demonstrated that the solution was required to allow free movement of the thumb such that it could form a pinch grip with the lateral side of the index finger and also un-obstruct the palm side of the hand such that it did not interfere with the object being grasped. Further to this, the index finger was required to be restrained during extension of the wrist. Without the restraint, the fingers extend outwards during wrist extension causing the thumb to ‘miss’ the lateral side of the index finger.

The required torque to hold the hand in an extended position was calculated by considering; the weight of the hand, weight of an object, and key pinch force. The weight of the hand was 600grams and was estimated using anthropometric data for 50% percentile male. The centre of gravity was approximated by the length between the centre of rotation of the wrist and the knuckle of the third finger. The length of 76mm was found using the Humanscale data sheet for 50% percentile male. The weight of a ‘heavy’ object as specified by the GRT was the videotape (3.5N). The length at which it acted was estimated at 100mm. The device aimed to provide a 10N pinch force which was specified as a functional force to achieve activities of daily living. (Smaby N 2004 et al). The three forces were combined to find the required maximum torque necessary to extend the hand about the wrist. The centre of rotation of the wrist was approximated as the head of the capitate carpel bone (Jackson WT 1994 et al). This resulted in a required maximum torque of 1.3Nm inclusive of 70% efficiency. The solution was required to provide 30° wrist extension, and 30° flexion to open the thumb. The combined range of motion of 60° was necessary for the thumb tenodesis to function adequately (Rothwell, 2005).

3.1.2 Activity

An activity was defined as the execution of a task or action by an individual. Examples of this are communication, mobility and self care. In most cases, these activities rely on the level of body functions and structures. Hanging out clothes for example, is only possible with the use of a pinch grip to pick up the washing; hence the activity of hanging out the washing is dependent on the ability to perform a pinch grip. Not all activities however are reliant on the level of body functions and structures. For example, mobility can easily be achieved with the use of an electric wheelchair, and does not require a pinch grip to perform this activity. Finally, the opposite result can also be achieved where the person may be provided by a pinch grip, but this solution actually prohibits performing certain activities. An example of this is a device that provides a pinch grip to hold a telephone; however the device is too heavy for the user to be able to bring the phone to their ear.

The assessment used to measure a person's activities with C5 tetraplegia was the Functional Independence Measure (FIM). This assessment evaluates the person's ability to perform tasks within the categories of; Self-Care, Toileting, Mobility, Communication and Thought Processes, Appendix A, P107. It uses a scale between 7-1 for each task, 7 being complete independence and 1 resulting in total assistance. A study carried out by the Mechanical Engineering Department, University of British Columbia interviewed potential users for an upper limb orthosis and asked them to identify a set of high priority tasks. The top 6 responses were; Reaching / picking objects, Personal Hygiene, Hobbies/Crafts, Eating / Drinking, Housework, and Dressing. These tasks are almost identical to the tasks associated with the Self-Care category of the FIM. Self-Care includes; Feeding, Grooming, Bathing, Dressing, and Toileting.

The same study researched the motion analysis to obtain data on the arm motions involved in performing these tasks. Figure 3-2 compares two various axis definitions, axis (a) is the standard medical terminology while (b) is more practical in assistive device design.

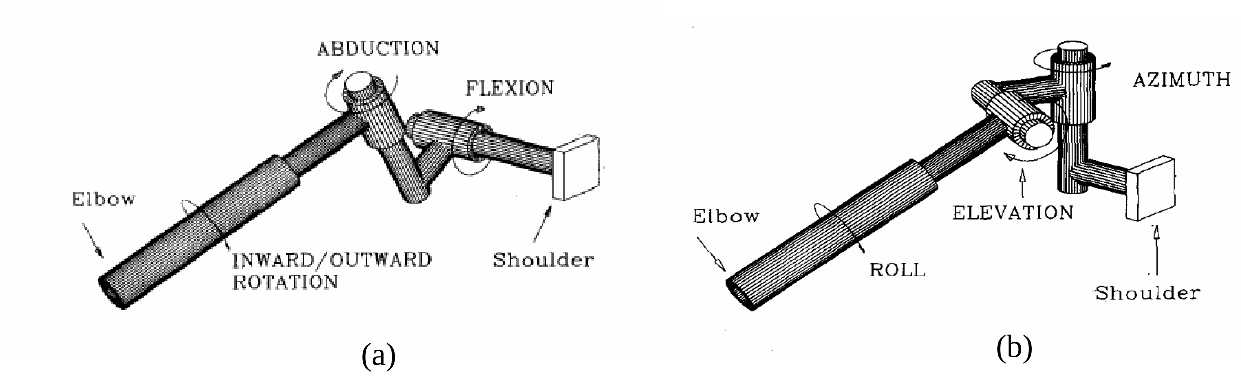


Figure 3-2 (a) Flexion-abduction rotation definitions. (b) Azimuth elevation roll definitions (Romilly et al., 1994)

In total, the ranges of motion of 22 ADL were measured using six able-bodied subjects. The subjects were instructed to perform the tasks as naturally as possible. The subject's trunks were constrained from moving forward while performing these tasks since the users of this solution would be unable to bend forwards due to the non-functioning abdominal muscles. Figure 3-3 shows the motion analysis results for eating with the hands. This shows some interesting results particularly with the relatively little movement of the Azimuth and Elevation compared the Elbow Flexion, Forearm Rotation, and Roll. This suggests that it is important for a solution to allow for Elbow Flexion, Forearm Rotation and Roll as these are critical motions while eating food. The average range for all 22 tasks were; Azimuth (40°), Elevation (35°), Elbow Flexion (39°), Forearm Rotation (52°) Wrist Flexion (33°), and Wrist Yaw (21°).

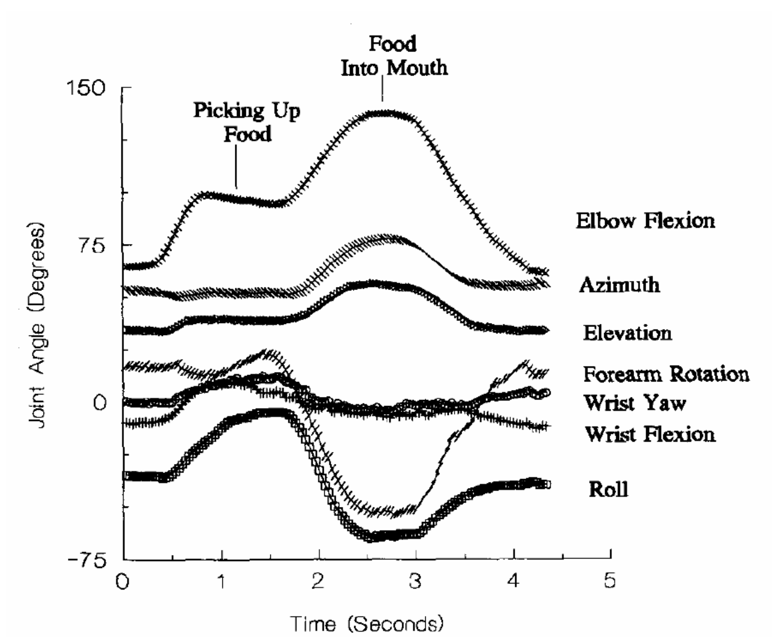


Figure 3-3 Motion analysis of eating with the hands definitions (Romilly et al., 1994)

3.1.3 Participation

The third factor, Participation, is defined as the involvement in a life situation. It is difficult to distinguish between Activities and Participation, but simply put; activity limitation is the difficulty in *execution*, while participation limitation is the difficulty with *involvement*. Participation involvement can be affected by stigmatisms, discrimination, or personal discomfort. It has been suggested that participation outcomes become critical whenever community based outcomes are assessed (Dijkers et al).

The Impact on Participation and Autonomy (IPA) questionnaire was used to identify key aspects of participation (Appendix A). The questionnaire is phrased as “My chances of getting around in my house where I want to are…..?” with 5 point answers between ‘very good’ and ‘very poor’. The questions cover a range of categories including: Mobility, Self-Care, Activities around the house, looking after money, leisure, social life, paid or voluntary work, education, and training. Though these questions are similar to the questions in 3.1.2 Activity, they are in fact different as participation is difficulty with involvement rather then difficulty in execution. Table 1 demonstrates key questions in the IPA that were considered important requirements for the solution design.

Table 1 Table of requirements found from the IPA questionaire

IPA Question	Product Requirement
My chances of going on the sort of trips and holidays I want to are	Good Reliablility (no batteries)
	Robustness (water resistant)
My chances of getting washed and dressed the way I wish are	Good functionality
My chances of eating and drinking when I want to are	Easy to put on and take off
My chances of using leisure time the way I want to are	Minimal time necessary to maintain the device
My chances of talking to people close to me on equal terms are	Good Aesthetics to avoid attention
My chances of having an intimate relationship are	Ease of use
	Quiet operation
My chances of getting or keeping a paid or voluntary job that I would like to do are	Simple device to reduce stigmatism

3.1.4 Environmental

Environmental factors make up the physical, social and attitudinal environment in which people live and conduct their lives. These factors are external to individuals and can have a positive or negative influence on the individual’s performance as a member of society, on the individual’s capacity to execute actions or tasks, or on the individual’s body function or structure. Environmental factors interact with the components of Body Functions and Structures, Activities and Participation. An environment without facilitators will restrict an individuals performance (e.g. inaccessible buildings) while environments which facilitate may increase performance (e.g. assistive mobility controls).

Environmental factors change the acceptability for an assistive device because it affects the need to have it. For example, if wheelchair ramps did not exist then wheelchairs would become obsolete. The same principles exist for a solution which improves grasping abilities. Environmental factors such as personal care and adaptive equipment all reduce the need for the solution. However if no personal care existed, then there would be a considerable need for an assistive solution. The current disabilities funding in New Zealand provides someone with tetraplegia with a care taker who in the morning and evening come to their house to support them. Combined with this, they have friends and family who also provide support. This network allows someone with Tetraplegia to function sufficiently in society which makes it very difficult to introduce a new solution as the person has already accepted their current environment. In fact, for an assistive device to become acceptable, it would have to impress the user such that they will voluntarily change their environment and choose the solution.

3.1.5 Personal

Personal factors, the second of the two contextual factors, are the particular background of an individual's life and living. This factor includes gender, race, age, other health conditions, and social background. Personal factors may have an impact on the outcome of various interventions. For example, if a person with tetraplegia has young children it will be difficult for them to have forearm tendon transfer surgery (FTTS) due to the relatively long rehabilitation period and time away from their family. Therefore they may prefer to accept a solution which requires less time to administer. Another example of a factor which significantly contributes to acceptability is gender. The differences of perceptions between males and females contribute significantly to the acceptability of a mechanical orthosis. Males for example may have less regard for their appearance then females, therefore a male may accept to wear a mechanical orthosis while the female would not.

3.2 The Design Requirements

From the analysis of Sections 3.1.1 through to 3.1.5 the following requirements listed in Table 2 were identified. The *Considerations* column identifies important features for the solution which were extracted from a range of sources including: physiotherapists; users; engineers; specialist surgeons; users; and analysis of the ICF. Each consideration was given a weighting of either a *Demand* or a *Wish* as shown in the *D/W* column in Table 2. Demands were considered essential to the solutions while the *wishes* were not.

Table 2 Design Requirement Specifications

DESIGN REQUIREMENT SPECIFICATION			
FACTORS	**CRITERIA**	**D/W**	**ACTIONED**
BODY FUNCTIONS	Maximum weight 200grams on forearm and wrist	W	MKS
	Provide 30 degrees wrist flexion	D	AR
	Provide 30 degrees wrist extension	D	AR
	10N Pinch Force	D	IKPF
	1Nm - 2Nm of wrist torque	D	MKS
	Operate at 60 RPM	W	MKS
	Free movement of the thumb	D	MKS
	Fixed finger position to provide lateral support for the key pinch	D	MKS
	Controllable	D	ICF
	Accurate Control of wrist position in both Extension and Flexion	W	MKS
	Palm of hand left un-obstructed	W	MKS
ACTIVITIES			
	First 35° of shoulder elevation remains unrestricted	W	FTA
	Auxiliary hand remains free of restrictions	D	R&M
	Natural wrist motion	D	MKS
	Adduction and Abduction wrist control	D	JD
	Mobile on wheelchair	D	JD
	Mobile on body	W	MKS
	Responsive	D	MKS
PARTICIPATION			
	Safe	D	ATOMS
	Quiet	W	ICF
	Hidden under clothing	W	ICF
	Unassisted fitting and unfitting	D	ICF
	Soft materials	D	MKS
	Breathable	D	JD
	Durable	D	MKS
	Robust and Waterproof	W	MKS
	1 min fitting time on and off user	W	ATOMS, MKS
ENVIRONMENTAL			
	One size fits all	W	MKS
	Common engineering materials	W	MKS
	Provide training	D	ATOMS
	Superior to competition devices	D	MKS
PERSONAL			
	Simple techonology	W	MKS
	Short training time 1-2days	W	ICF

KEY	
MKS	Mathew Kyle Singer
ICF	International Classification of Disability and Health
IKPF	Identification of key pinch forces (Smaby et al., 2004)
R&M	Reaching and Manipulation
FTA	Functional Task analysis P124 (Romilly et al., 1994)
ATOMS	Assistive Technology Outcomes measurement System (Lauer et al., 2006)
AR	Professor Alastair G Rothwell
JD	Jennifer Dunn (Physio Therapist for Burwood Spinal Cord Unit)

Chapter 4 Concept Generation

The concept generation process was necessary to transform the solution into a concept. The final aim was to create a concept which fulfilled the design requirements listed in Table 2. To complete this, the concept was split into three sections which were fundamental to the concept, this included: *energy* to power the concept; a *mechanism* to transform the energy into motion; and *control* to make the concept useable. These functions each had a number of alternative solutions as demonstrated in Figure 4-1 through to Figure 4-4.

4.1.1 Energy

The design requirement specifications identified an energy source which was mobile; light; safe; quiet; and relatively inexpensive. Shown below in Figure 4-1 are four different energy solutions which represent four unique options to drive the mechanism. Though there are many more alternative energy sources, the ones listed below represent the different *forms* of energy, while different energy *sources* are investigated during the sub-system design.

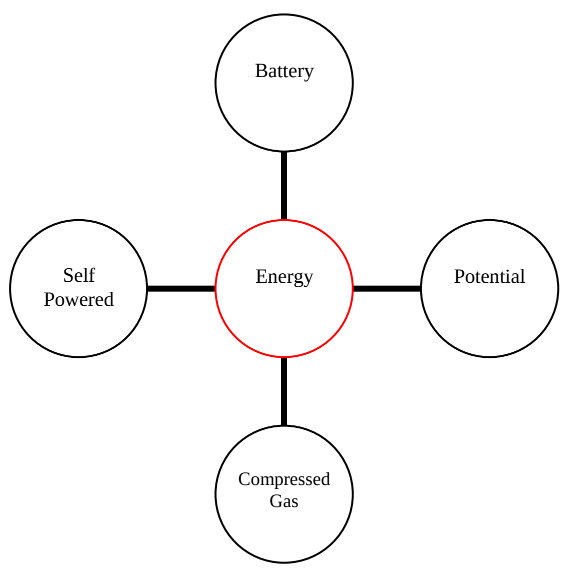


Figure 4-1 possible energy solutions

Two locations were considered suitable to place the energy source as defined in the design requirements; this was either on the wheelchair or attached to the body. Housing the energy source on the wheelchair provided several key advantages which included decreased weight on the body and improved aesthetics. The disadvantage was that the user was ‘plugged’ into the wheelchair which was considered impractical as outlined in the design requirement specifications. The second location to house the energy source was on the body, however difficulties included weight; safety; and noise.

4.1.2 Mechanism

The second input function for the design was the mechanism which was used to transform the energy into a useable force. Six solutions were considered and are shown below in Figure 4-2. Though these are only a sample of different mechanism options, they represent a wide variety of technology which is the aim of the concept development.

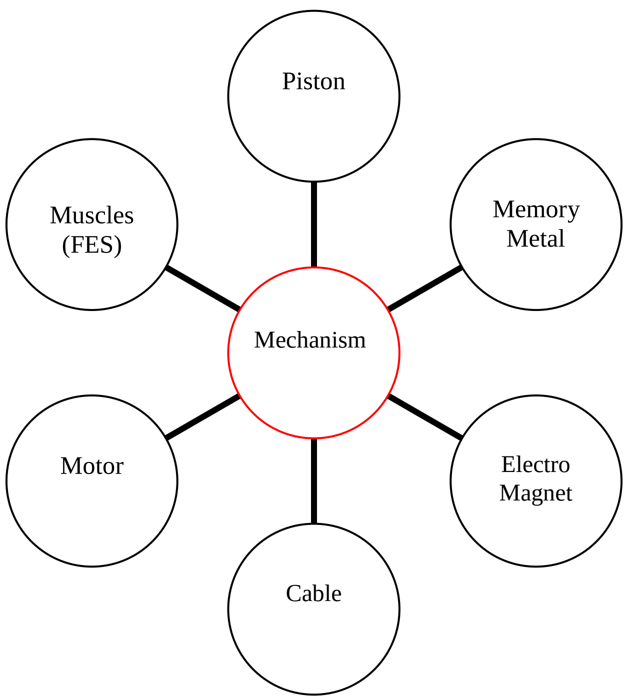


Figure 4-2 alternative function components for the design conceptualization

4.1.3 Control

Control of the hand is a critical function during grasping activities because the arm functions principally to place the hand in the appropriate position, while the hand's major purpose is to interact with the environment. (Carr and Shepard, 1980). Figure 4-3 shows the range of wrist positions required whilst performing ADL, this data suggests that each task requires a different range of wrist positions. Therefore a control system which simply switched between a set range of angles over a set time was considered impractical. The control system for this device was required to control the angle of the wrist, and the rate of change of the angles as defined in the design requirements.

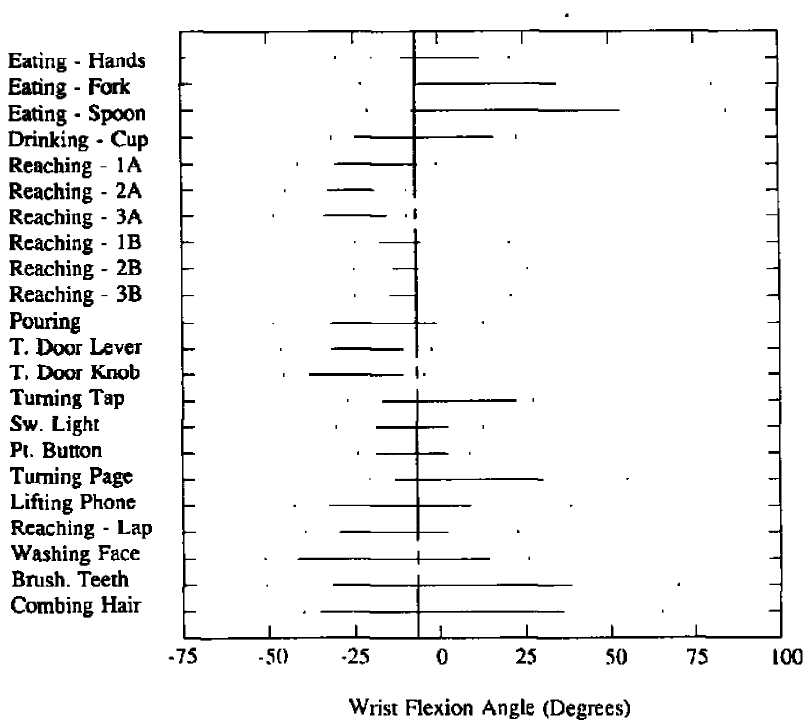


Figure 4-3 Wrist Flexion, average min/max & extremes (Romilly et al., 1994)

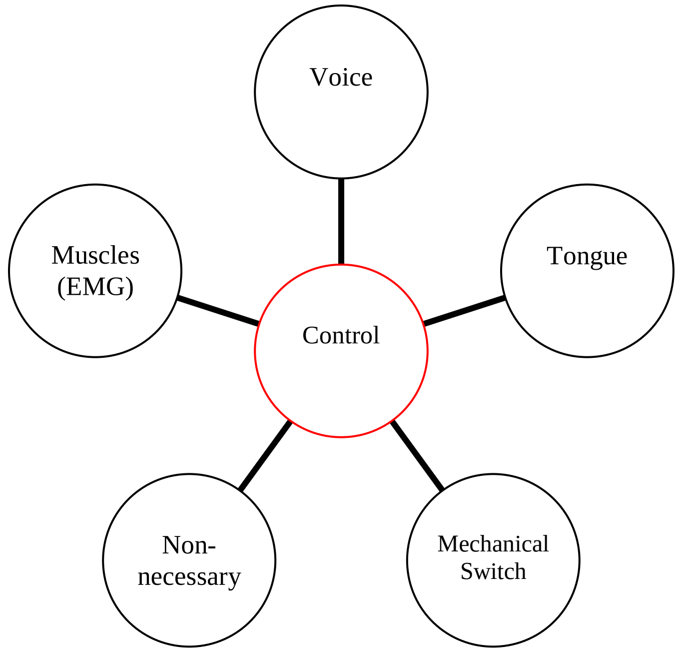


Figure 4-4 Mind map of various control alternatives

Figure 4-4 shows various control solutions for use with the concept. Voice and Tongue control were both considered impractical as they may impair on social acceptance.

Myoelectric control records the potential difference generated during a muscle contraction and requires sensors to be accurately positioned over functioning muscles. One of the most significant advantages of myoelectric control signals is that they can be obtained from the action potentials of contracting muscle fibers despite little or no movement, making the device attractive for patients with weak or small muscles (Benjuya and Kenney, 1990). However disadvantages include complexity of use, reliability of control, and cosmetic appearance.

The mechanical switch is a simple device which requires manipulation of a control mechanism. For example the auxiliary forearm could be used to operate a switch which in turn would change the wrist position. The disadvantage of this concept is that most commonly performed tasks are carried out not with one hand but with two (Carr and Shepard, 1980). Therefore this concept may in fact impair their performance. The alternative position for the switch was to utilize the auxiliary shoulder by placing a control mechanism just above the pectoralis muscle. The control mechanism would measure elevation or depression of their shoulder which in turn controlled their wrist position. The advantage of this control mechanism is the ability for fine control and adjustment.

The final control option was to completely avoid a control mechanism through such means as running a cable between the auxiliary shoulder and tricep muscle. This would be a similar design used with the body powered prosthetic limbs which functions with a combination of auxiliary shoulder movement and arm elevation.

4.2 Concept Construction

The Table of Options, Figure 4-5, was used to develop four unique concepts; the concepts were developed by selecting one solution from each row. Each concept was chosen for its unique qualities and characteristics. This method of concept development was extracted from a systematic engineering design approach (Wallace and Gooch, 2004).

Function \ Solutions	1	2	3	4	5	6
Energy	Potential	Compressed Gas	Battery	Self-Powered		
Mechanism	Piston	Muscles (FES)	Electro-Magnet	Motor	Memory Metal Nitinol	Self-Moved
Control	Muscles (EMG)	Physical	Voice	Tongue	Sensors	Self-Controlled

Muscle Excitation | Gas Piston | Electric Motor | Self Powered

Figure 4-5 Table of options used to develop four concepts

In practice, some of these concepts would take a long time to develop and some would cost much more then others, however, the reason for constructing these four concepts was to evaluate them based on their hypothetical outcomes. Sections 4.2.1 through to 4.2.4 show a simplistic version of each concept.

4.2.1 Concept One Muscle Excitation

Electrodes planted on the surface of the skin excite the extensor muscles to produce extension at the wrist. The auxiliary shoulder is used to activate and control the motion.

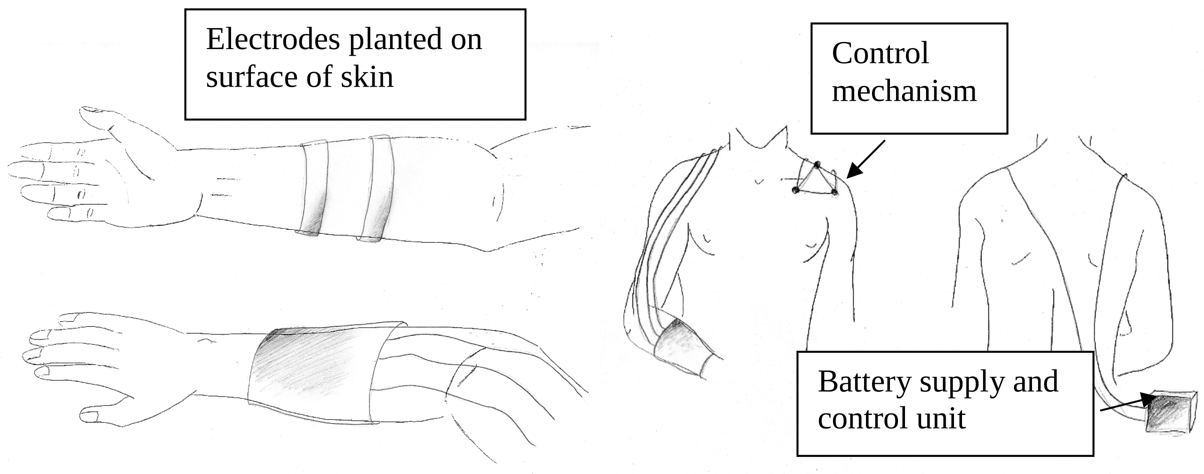


Figure 4-6 Electrical Muscle Stimulation

4.2.2 Concept Two Gas Piston

The hand brace is connected to a cable which runs to a gas piston located on the wheelchair. Activation is controlled by the auxiliary shoulder which in turn controls the gas piston.

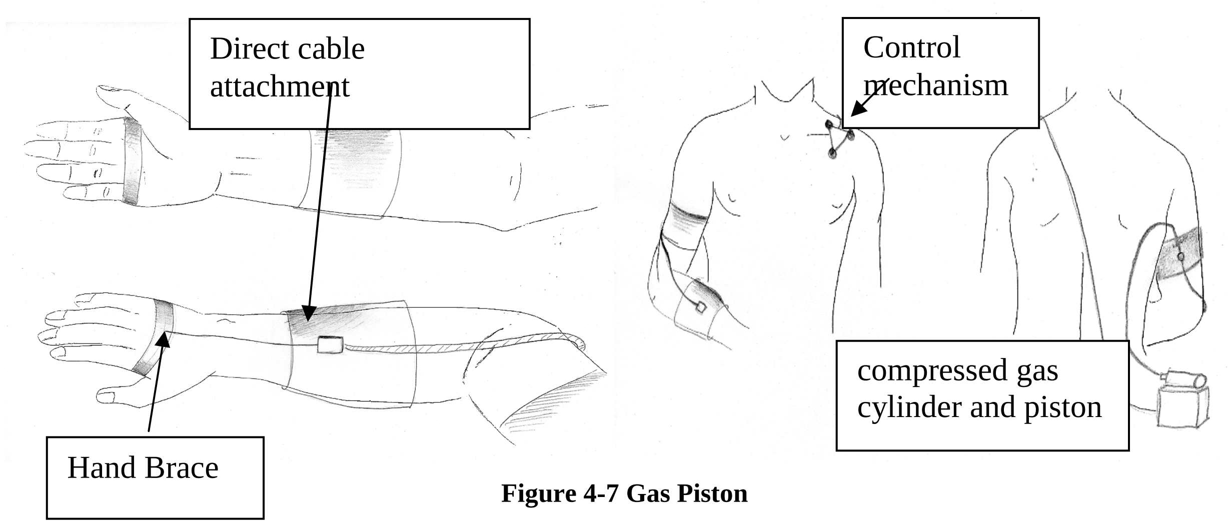


Figure 4-7 Gas Piston

4.2.3 *Concept Three Electric Motor*

An electric motor is attached to the forearm and through a linkage bar it pushes the hand into extension. Activation is controlled by the auxiliary shoulder which in turn controls the power to the motor.

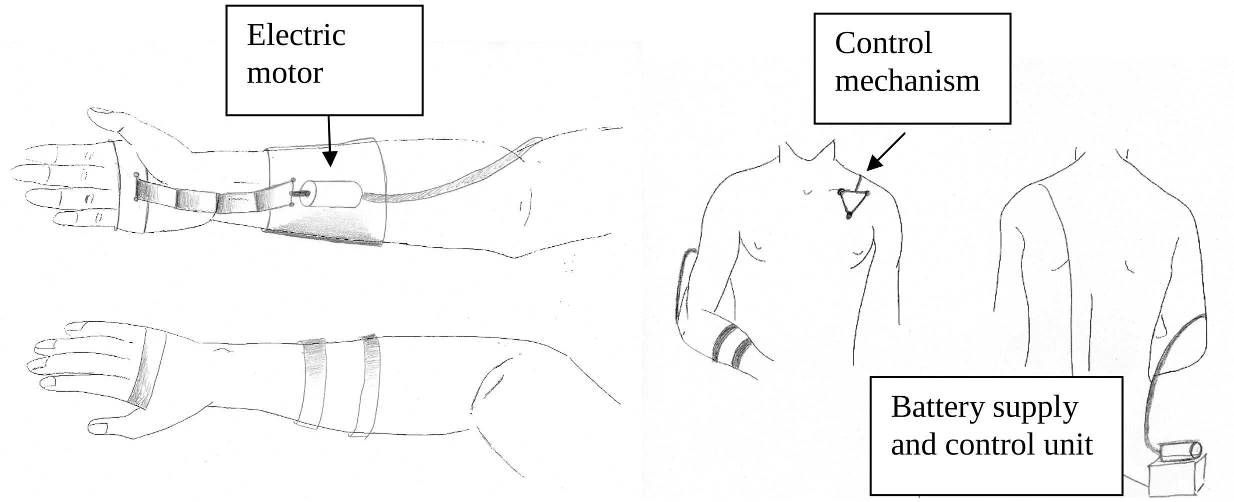


Figure 4-8 Forearm Gas Piston orthosis

4.2.4 *Concept Four Self Powered Wrist Extension Orthosis*

The concept utilizes the shoulders to power the orthosis. These motions combined provide tension to a cable which in turn extends the wrist. Control and activation is dependant on the position of the shoulders.

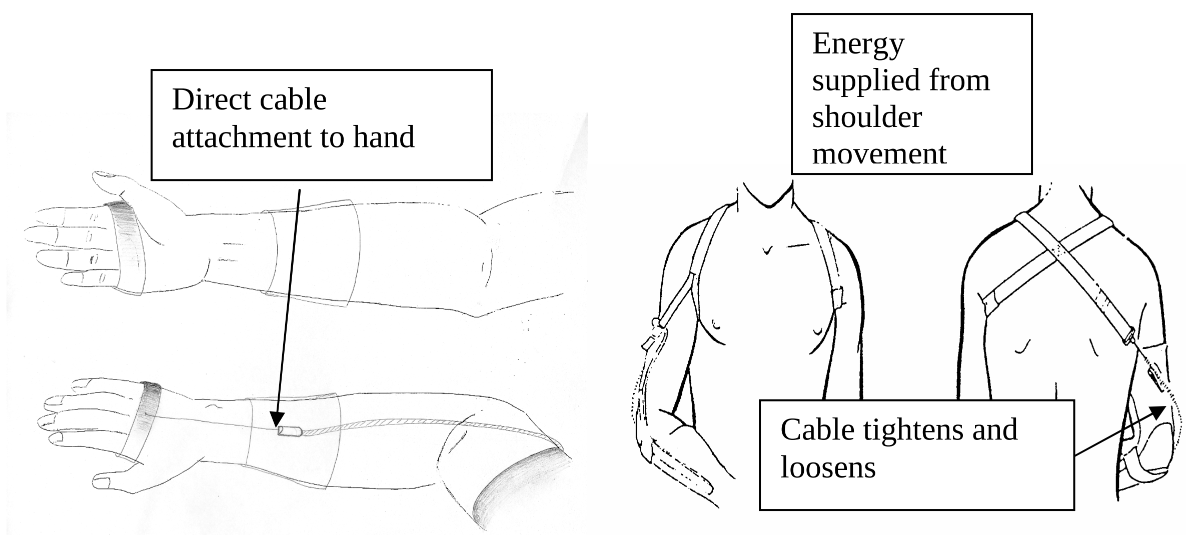


Figure 4-9 Auxiliary Shoulder orthosis

4.3 Concept Evaluation

A Concept Evaluation Chart, Table 3 (Wallace and Gooch, 2004) was used to determine the optimal concept from the four identified concepts found in Section 4.2. The left hand column of the concept evaluation chart contains the criterion which was used to compare between the four selected concepts. Each criterion was given a weighting which was dependent on its importance to the overall concept. The *forearm gas piston* concept was chosen as the datum and was given a score of zero for each criterion, each of the subsequent concepts were given a score relative to this datum. Each score was multiplied by the weighting (Column 2) to give the *weighted (Wt) Value*. The weighted scores were combined and the concept with the highest score was theoretically the optimal concept.

The criteria in the concept evaluation chart was selected from the ‘Wishes’ identified in the design requirements specification checklist as shown in Table 2, P30. The ‘Demands’ were not used in the evaluation criteria as hypothetically each concept should fulfill the ‘Demands’ identified in the design requirements. Adding the ‘Demands’ to the evaluation chart would result in the same score between each concept. However the ‘Wishes’ demonstrates subtle characteristics between alternative concepts and therefore was chosen as the comparison criteria (Wallace and Gooch, 2004).

Table 3 Concept Evaluation Chart

CRITERIA		Electric motor	Self Powered Wrist Extension Orthosis		Muscle Excitation		Gas Piston	
	Weighting		Value	Wt Value	Value	Wt Value	Value	Wt Value
Weight on the Arm	3	Datum	2	6	0	0	0	0
Operation Speed	1		-1	-1	-1	-1	1	1
Accurate Control	3		-1	-3	1	3	0	0
Unobstructed Palm	2		0	0	1	2	0	0
Unrestricted Shoulder Elevation	2		-2	-4	0	0	0	0
Mobile on Body	1		2	2	0	0	0	0
Quiet	3		2	6	2	6	0	0
Hidden Under Clothing	1		2	2	1	1	0	0
Robust and Waterproof	1		2	2	-1	-1	0	0
1 min Fitting Time	2		-1	-2	-1	-2	0	0
Low Cost	1		2	2	-2	-2	0	0
Common Engineering Materials	1		0	0	-2	-2	0	0
Simple Technology	2		2	4	-2	-4	0	0
Short Training Time	2		1	2	-2	-4	0	0
TOTAL SCORE		0		16		-4		1

4.4 Selected Concept

The *self powered wrist extension orthosis*, referred to as the *orthosis*, was chosen as the preferred concept because this hypothetical solution excelled over the other three concepts. The orthosis used a harness which extended across the user's shoulders; as the user manipulated their shoulders a cable extended and curtailed which in turn controlled the wrist position.

A key pinch would be achieved through a simple operation which anchors a thumb tendon to the radial bone. As the wrist extends, the anchored tendon pulls the thumb firmly towards the index finger producing a key. Combining both surgery and the orthosis together was proposed by Professor Alastair G Rothwell, Head of Department of the Christchurch School of Medicine. Detailed information of the operation is given in Section 2.3.1, Surgical Solutions.

Appendix H, P165 shows further examples of constructed concepts which were used in this research.

Chapter 5 Sub System Design

The concept generation proved that the *self powered wrist extension orthosis* provided a simple, controllable, and useable means to provide energy to the wrist. The energy was supplied through a cable which moved a set distance with a set force and was dependant on the strength of the user's shoulders. The aim for the sub system design was to determine the optimal method for converting this energy into wrist motion given the physical contraints of the user's shoulders. The concept was separated into two sub-systems which were analyzed separately. The first sub-system was called the *shoulder harness* and the second was the *wrist harness*. Solutions were found for each sub-system and combined to complete the overall orthosis.

5.1 Shoulder Harness

Figure 5-1 shows the 'figure-of-eight' shoulder harness system used for many years in prosthetic limb design. The success of the harness can be contributed to the low cost, high reliability, light weight and simplicity, combined with the kinesthetic[4] feedback provided by the harness system. The harness provides energy through a cable which moves a set distance (stroke), with a set force, and both are dependant on the strength and range of motion of the user's shoulders. The stroke and force were both related to the change in length between the neutral position and the elevated position (Abduction or flexion) as seen in Figure 5-2. Combined with this the auxiliary shoulder also contributed to the stroke and force through protraction and retraction.

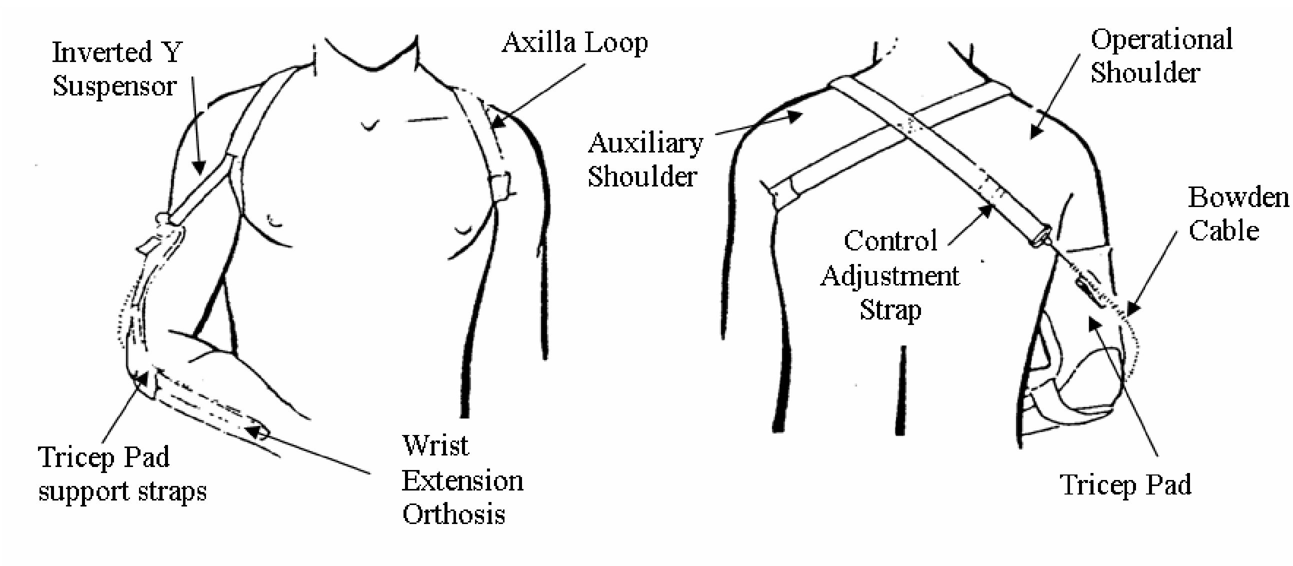


Figure 5-1 The shoulder harness definitions

[4] The sense that detects bodily position, weight, or movement of the muscles, tendons, and joints.

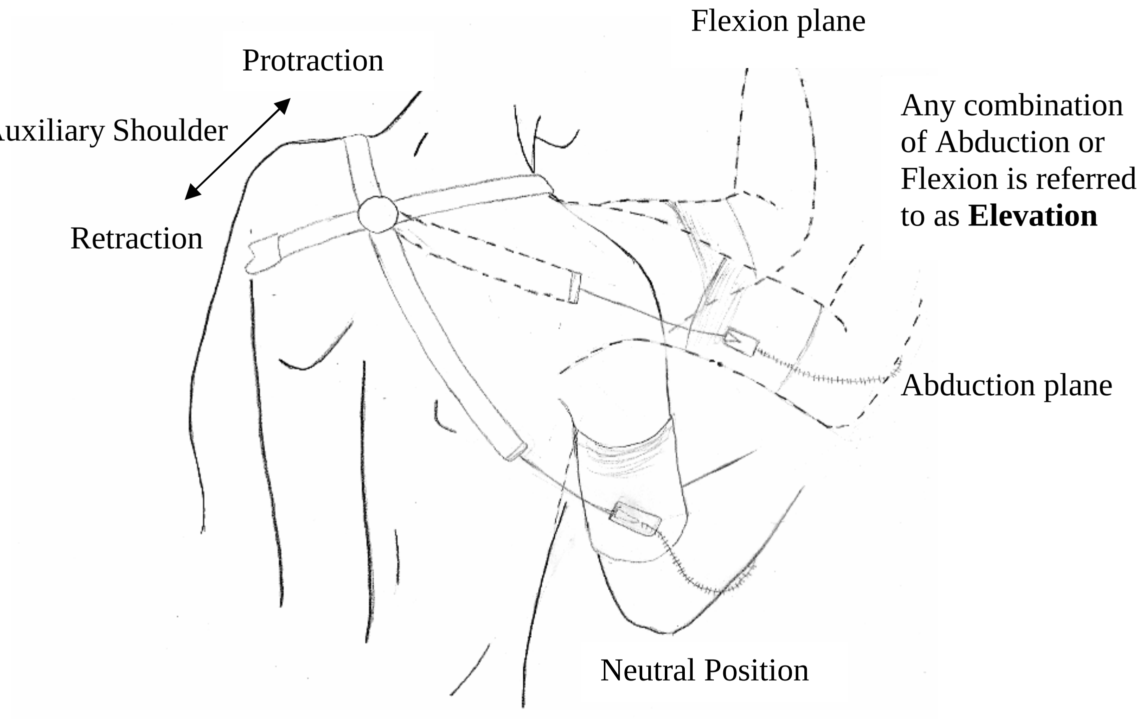


Figure 5-2 Schematic of Bowden cable and arm position

5.2 *Wrist Harness*

The sub-system design of the wrist harness was split into three smaller systems which included; wrist kinematics and dynamics; forearm brace design; and hand brace design.

5.2.1 *Wrist Kinematics and Dynamics*

The aim of the wrist function analysis was to determine the optimal method for controlling the wrist, given that the energy was to come from a single cable which extended and curtailed. The most basic method to provide wrist extension was to attach a cable directly to the posterior side of the hand i.e. the knuckles, such that as the cable tightened the hand pulled backwards into extension. This solution functions well as long as the cable is always in tension, however, as seen in Figure 3-3, P26 forearm rotation is common during grasping activities, hence if the cable is compressed the system loses control. An improvement on this solution was to add a spring which is attached to the palmer side of the hand. The advantage of this is that the hand is always ‘pulled’ towards flexion, thus the hand will remained stable irrespective if the hand is turned upside down. This solution however still has one key problem, for example, if the person wished to write a letter, they require a key pinch to hold the pen, a key pinch however requires shoulder elevation to produce tension in the cable. The problem therefore is while the person is writing they are required to consciously elevate their shoulder to maintain tension in the cable; this is awkward and tiring and eventually the muscles holding the shoulder in elevation will fatigue.

This problem was resolved by reversing the system such that the spring held the wrist extension and the cable pulled the wrist into flexion as shown in Figure 5-3. The advantage of this was that the hand remained in extension providing a pinch grip at all times. When the user wished to open the grip they simply applied tension to the cable which flexed the hand. This system allowed the user to maintain a pinch grip without any effort, hence as they performed day-to-day activities they could concentrate on the tasks, and not on maintaining the pinch grip. Further advantages of this system were that as the user brings an item closer to their body the pinch grip increases in strength; while the further away the user reaches the weaker the pinch grip becomes. This design requires the wrist to be held in extension for long periods of time, however this was favorable as it prevents finger contractures (Rothwell, 2005).

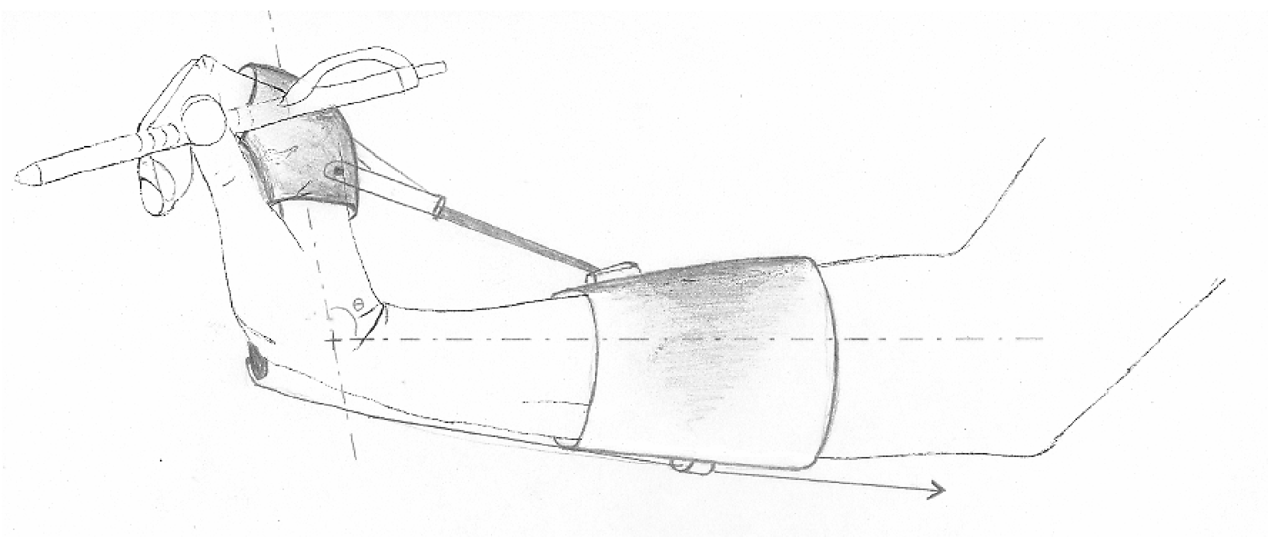


Figure 5-3 Sample of the wrist mechanism

5.2.2 Forearm Brace

The forearm brace was required to provide a platform for the cable and spring and transfer these forces back through the body structures. The forearm brace diagram, Figure 5-4, shows the forces acting on the system. Due to the dynamics of the design, the x-component spring forces were considerably large causing the arm brace to slide down the forearm. One method to improve was to increase the friction generated between the arm brace and skin of the forearm. However this would cause a number of problems including pressure sores and excessive sweating resulting in further skin irritation. The solution therefore was to utilize the tricep pad support straps (Figure 5-1) which supported the arm brace in the x direction by transferring the forces onto the tricep muscles. The y-direction forces were transferred onto the forearm through the arm brace and were considerably small. This design allowed for a relatively simple forearm brace which was easy to take on and off and applied minimal pressure to the forearm. The alternative solutions are shown in Appendix B, P111.

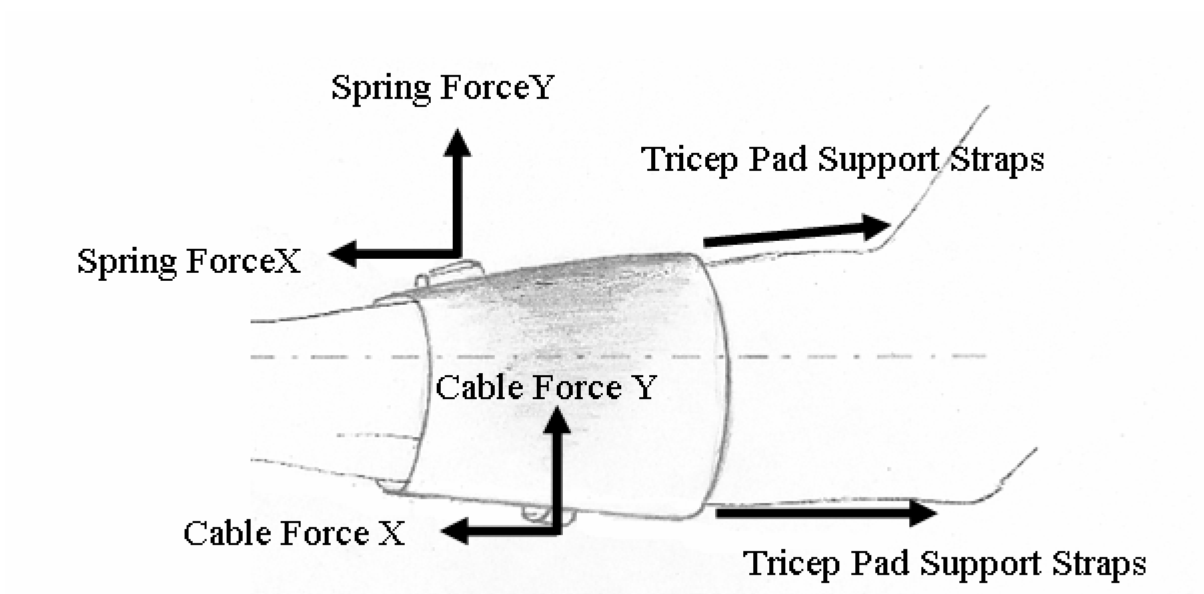


Figure 5-4 Free body diagram of the arm brace

5.2.3 Hand Brace

Critical design features of the hand brace were as follows:

- The hand brace was necessary to evenly transfer the forces from the spring and cables over the hand. Any concentration in pressure may result in pressure sores resulting in further hospitalization and rehabilitation. Figure 5-5 shows the forces while the hand is held in its resting position. Figure 5-6 shows the increased forces acting on the hand brace during flexion.

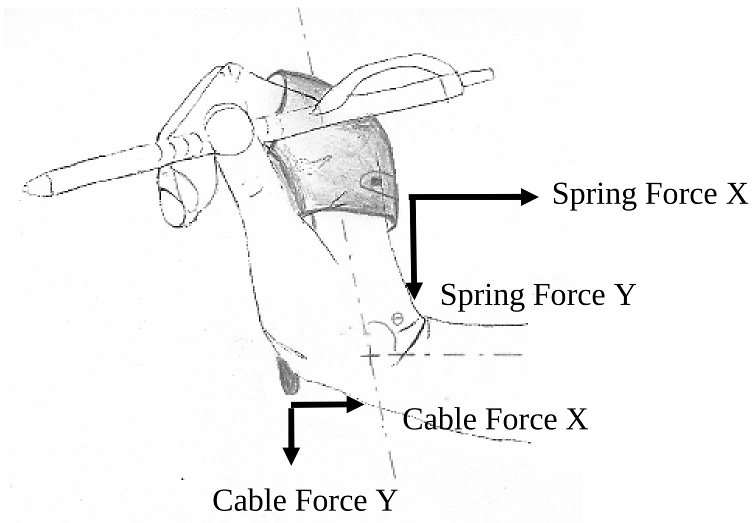


Figure 5-5 Forces on the hand while resting in extension

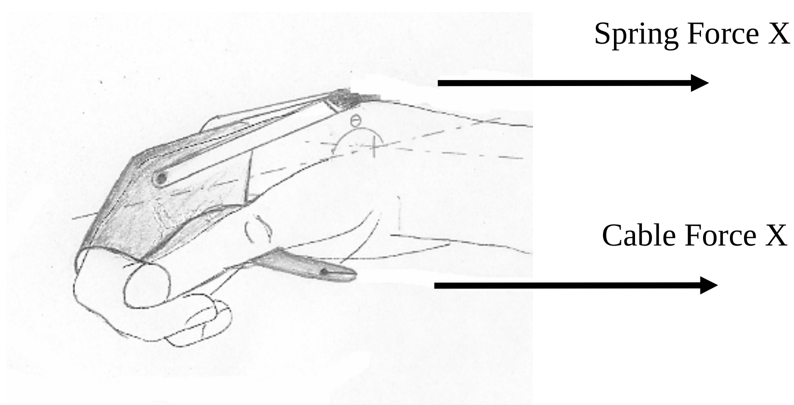


Figure 5-6 Forces on the hand while flexed forwards

- The hand brace was required to fit easily without the help from an assistant as defined in the design requirements specification.

- The hand brace was required to hold the hand in the correct hand posture so that the fingers supported a ‘key pinch’ grip.

- The ‘gate’ between the thumb and index finger was required to be left unobstructed so that the thumb could perform a natural pinch grip without impedance from the brace (Figure 5-7).

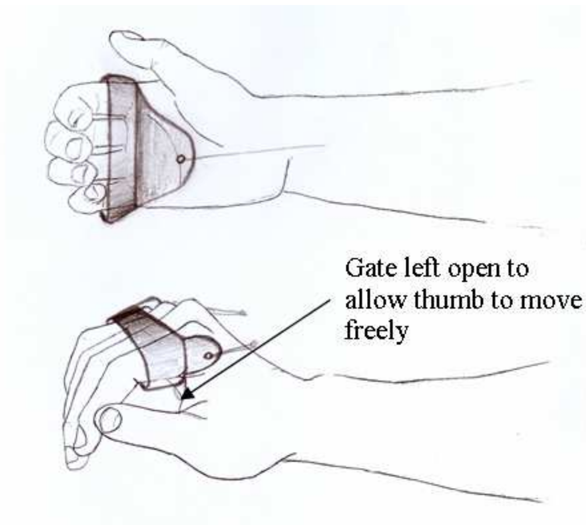


Figure 5-7 Final hand brace design

The result was a hand brace made from moldable plastic which was formed into a shape so that while the hand was held in extension, the x-component forces were distributed over the palm of the hand, and while the hand flexed forwards the x-component forces were distributed over the fingers and knuckles as shown in Figure 5-7. During user trials it was found that the moldable plastic hand brace distributed loads evenly over the hand; while hand braces made from fabric ‘sagged’ and produced points of high pressure. The results of the experimental hand brace designs are shown in Appendix B, P111. The Appendix identifies features of concern between various concepts.

The moldable plastic hand brace was customized to fit each user which reduced the need for straps, and allowed an assistant to slip it on or off the user. With further development, the aim was for the user to fit the moldable hand brace without help from an assistant. The hand brace also restrains the fingers from flexing outwards and provides a platform suitable for a key pinch grip; this was identified during user trials.

5.3 Final Concept Design

The final concept holds the wrist in an extended position such that a passive key pinch is always active. This allows the user to move their arms around freely while maintaining a key grip. It provides the user with maximum function as the arm is free while the user is holding an object. When the user wishes to flex the wrist, and hence drop the object, they do so by applying force to the cable as shown by the arrow direction in Figure 5-8.

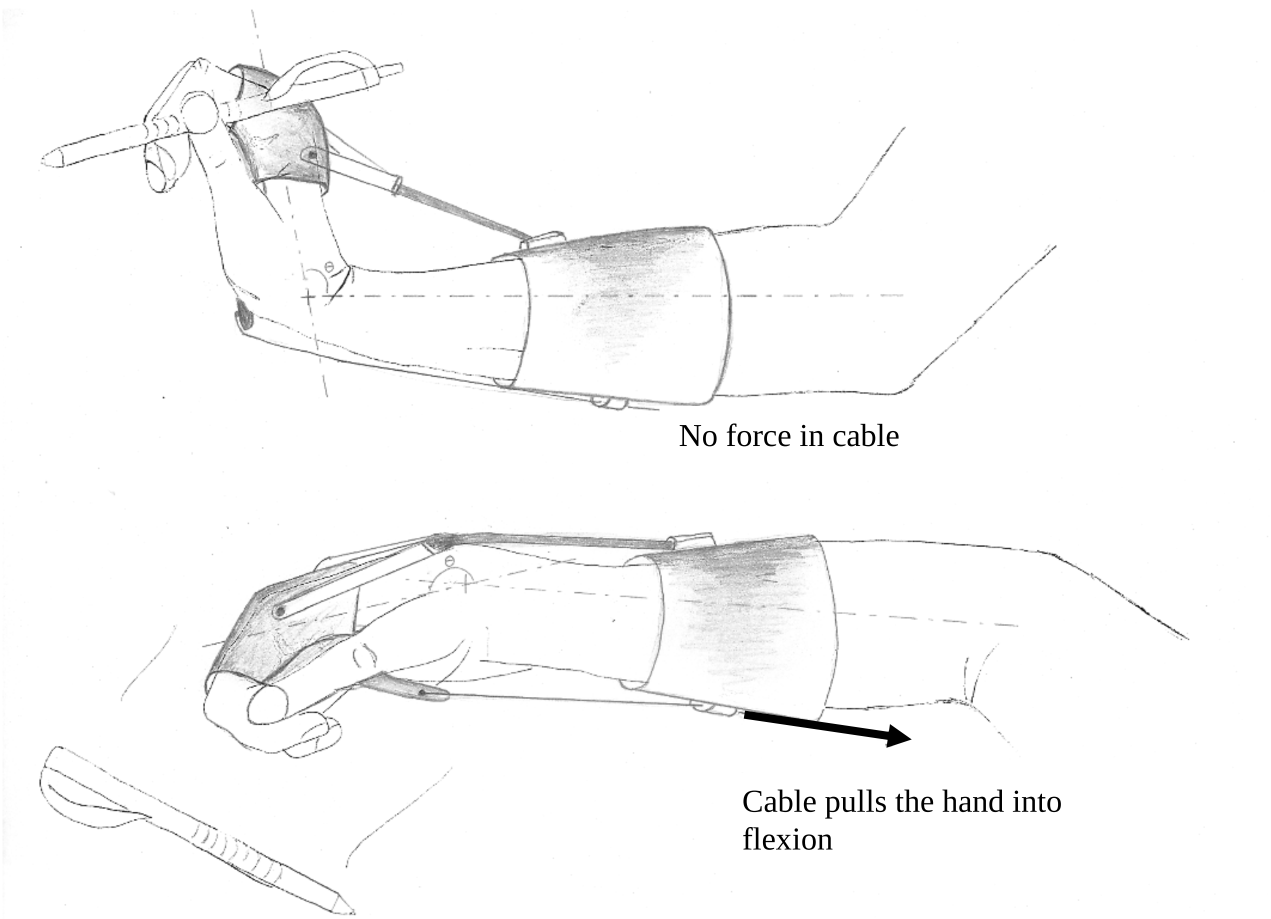


Figure 5-8 Final wrist harness design

The tension in the cable was controlled by the shoulder harness which is partially controlled by the auxiliary shoulder through the axilla loop and the operational shoulder. The auxiliary shoulder essentially provides fine adjustment to the cable tension through protraction and retraction while the main force is provided by elevation of the operational shoulder as described in Figure 5-2, P45.

Chapter 6 Concept Development

The *self powered wrist extension orthosis* was separated into two sub-systems which included the shoulder harness and wrist harness. The shoulder harness supplied energy to the wrist harness which in turn moved the wrist. The energy supplied by the shoulder harness was called the *achievable* energy whereas the energy used by the wrist harness was called the *required* energy. Through comparing the achievable energy supplied from the user's shoulders to the required energy necessary to pull the wrist from extension through to flexion, it was possible to pre-determine the functional outcome of the orthosis. The aim of the concept development was to improve the concept design so that the required energy did not exceed the achievable energy.

The analysis of the required and achievable energy sources were based on the fundamentals of body kinematics and dynamics. The dynamics of the upper body determined how the forces generated from the shoulder compared to the forces required to pull the wrist from extension through to flexion. The term used to describe the force generated by the shoulder became known as the *achievable force* while the necessary force to pull the hand from extension through to flexion became known as the *required force*. The kinematics determined the required change in cable length necessary to pull the wrist from 30° extension through to -30° extension, i.e. 30° flexion. This term was known as the *required stroke* and similarly the *achievable stroke* was from the output of the shoulders. For the remainder of the report +30° relates to wrist extension, while -30° relates to flexion.

Through analyzing both the achievable and required energies it was possible to predict the angle of wrist rotation where the required energy exceeded the achievable energy. This angle of wrist rotation was known as the *attainable wrist rotational angle* and this value was used as a datum to measure between different variations of the same orthosis.

6.1 Shoulder Dynamics and Kinematics

Figure 6-1 shows the definitions of parts of the orthosis, as well as the planes of motion (shoulder flexion and abduction), and axis definitions for the tricep pad. These definitions are used during the calculation of the achievable stroke and force.

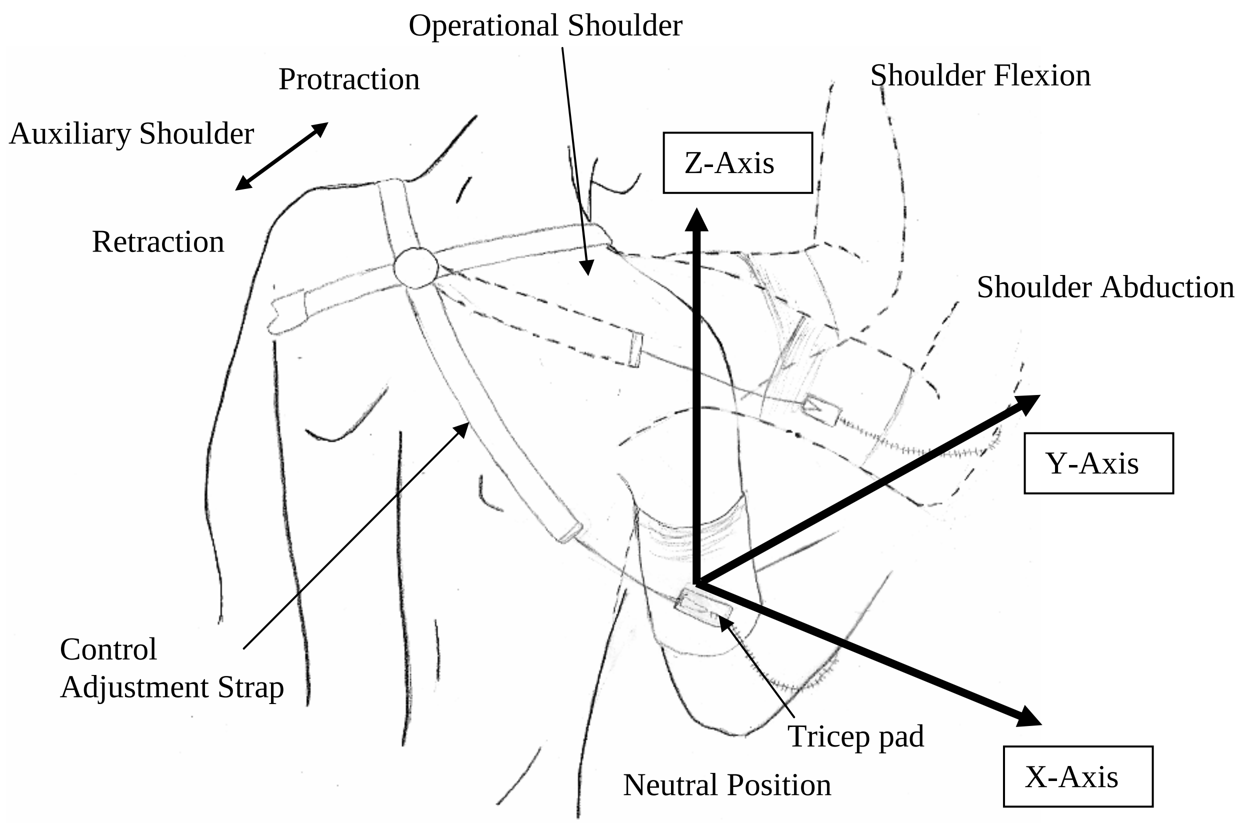


Figure 6-1 Shoulder Harness Characteristics

The key measurements necessary to determine the achievable stroke and force were the user's strength and range of motion (ROM) of the operational shoulder. After consultation with the head physiotherapist at the Burwood Spinal Cord Unit, two persons with tetraplegia were introduced and participated in testing of their shoulder strengths. Participant #1 was a 23 year male who had C4/C5 incomplete tetraplegia and was approximately 6 months post injury; participant #2 was a 26 year old male who had C5/C6 incomplete tetraplegia and was approximately 4 months post injury.

A hand held goniometer and force gauge (Hoggan Health MicroFet3) was used to measure each of the participant's shoulder ROM and shoulder strengths. The ROM was measured by recording the angle of the arm in the neutral position, and then at the maximum range in abduction and separately in flexion (Figure 6-2). The second sets of measurements were the shoulder strengths. Once again the MicroFet3 was used to measure the maximum achievable force in abduction and separately in flexion. Table 4 shows the experimental results from the two participants.

Table 4 Experimental shoulder strength results

Participant #1		Achieved Angle (°)	Maximum Force (N)
Shoulder Abduction	Trial 1	85	30
	Trial 2	84.5	26
	Average	**85**	**28**
Shoulder Flexion	Trial 1	48	37
	Trial 2	65	35
	Average	**56**	**36**
Participant #2		Achieved Angle (°)	Maximum Force (N)
Shoulder Abduction	Trial 1	103	105
	Trial 2	100	95
	Average	**101**	**100**
Shoulder Flexion	Trial 1	76	85
	Trial 2	77	87
	Average	**76**	**86**

The results show that Participant #2 was substantially stronger then participant #1 and for this reason the concept was based on the weaker participant who was also considered to have average/low shoulder strengths compared to other people with C4/C5 tetraplegia. Figure 6-2 shows the results from Participant #1. The figure shows that while the arm was in the neutral position the participant was able to produce 28N abduction and 36N flexion and the maximum angles were 85° abduction and 56° flexion.

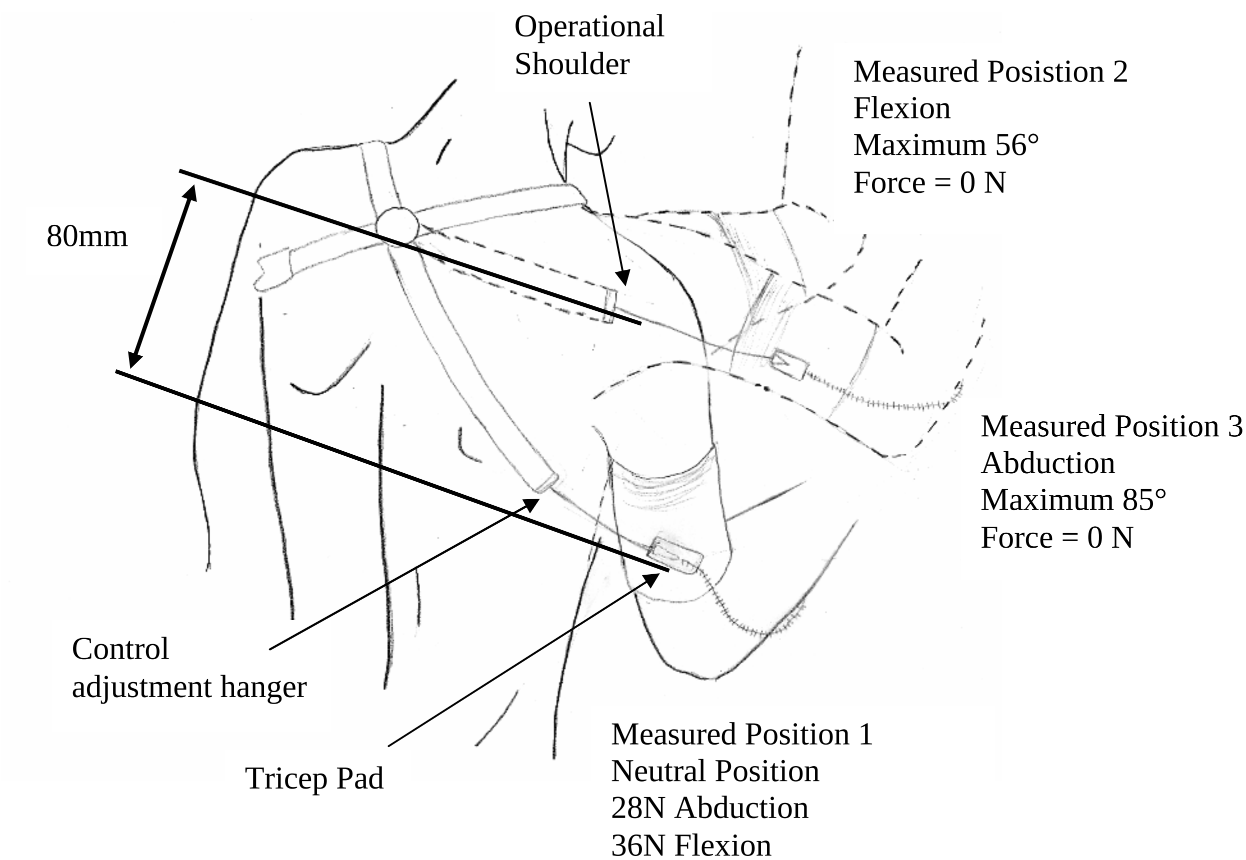


Figure 6-2 Important shoulder strengths to power the orthosis

6.1.1 Achievable Stroke Length

The achievable stroke length was calculated by firstly mapping out the possible positions of the tricep pad which was called the *tricep pad position map*. From this, the stroke length was calculated by comparing the position of the control adjustment hanger to the theoretical position of the tricep pad. The final graph of the *achievable stroke length* compares stroke length to wrist rotation. The information of the following pages outlines the procedure used to calculate the achievable stroke length while the Matlab code created to develop all of the following figures is shown in Appendix C, P117.

A *tricep pad position map*, which shows the possible positions of the tricep pad, was created using Matlab and is shown in Figure 6-3. The three measured positions were taken from Participant #1 and were the measured angles formed between the tricep pad and body (Table 4). These measured angles occurred about the shoulder joint and the distance between the tricep pad and shoulder joint was 80mm as shown in Figure 6-2. From this data it was possible to map all the possible positions of the tricep pad using Matlab. The origin, (0,0,0), was the neutral position of the tricep pad; the X-Z plane tracks the tricep pad in abduction while the Y-Z plane is flexion.

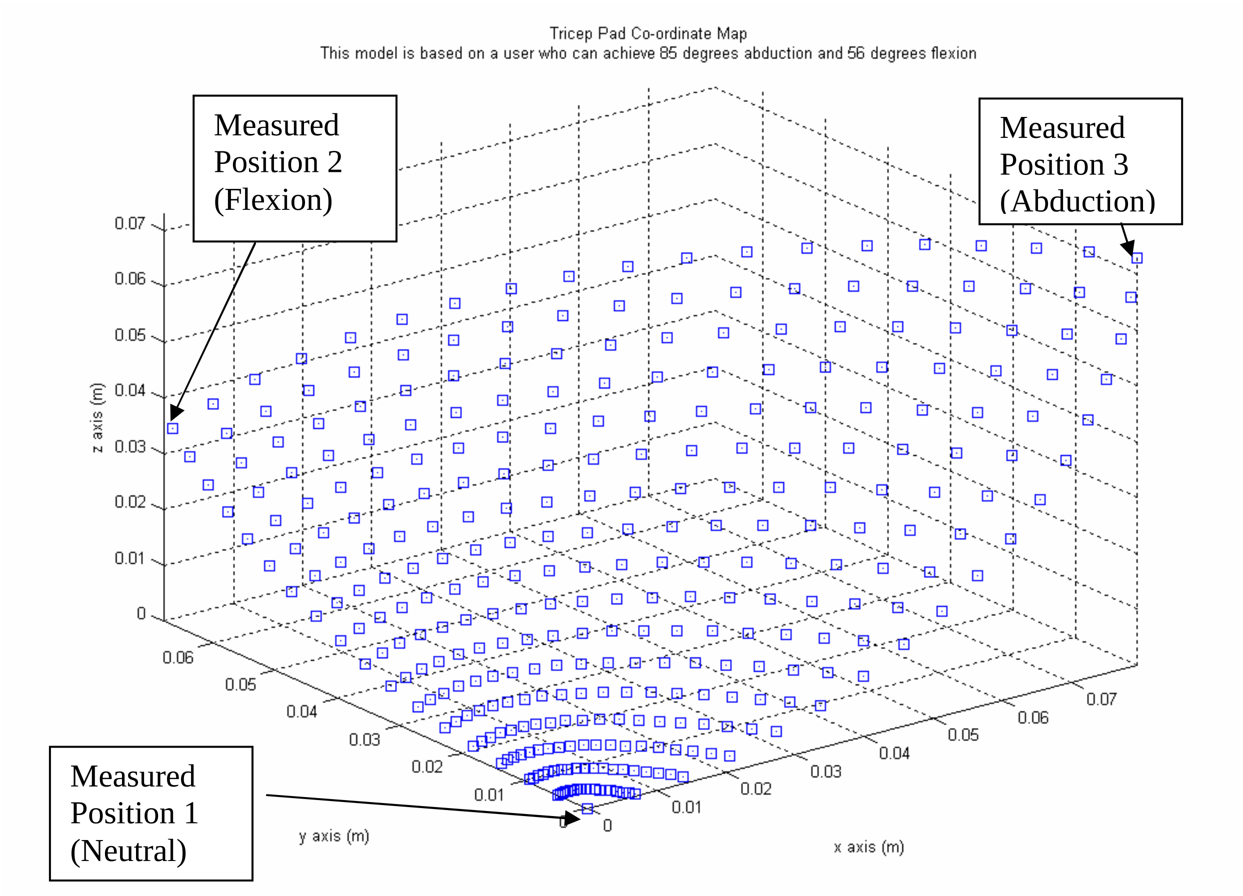


Figure 6-3 Tricep pad position map

The data obtained from the tricep position map was used to calculate the achievable stroke which was calculated based on the distance between the position of the tricep pad and the position of the control adjustment hanger as shown in Figure 6-2. Two assumptions were made about the control adjustment hanger: during shoulder abduction, the control adjustment hanger changed position; however during flexion it tightened against the body and held its position. Figure 6-4 shows how the stroke length was calculated.

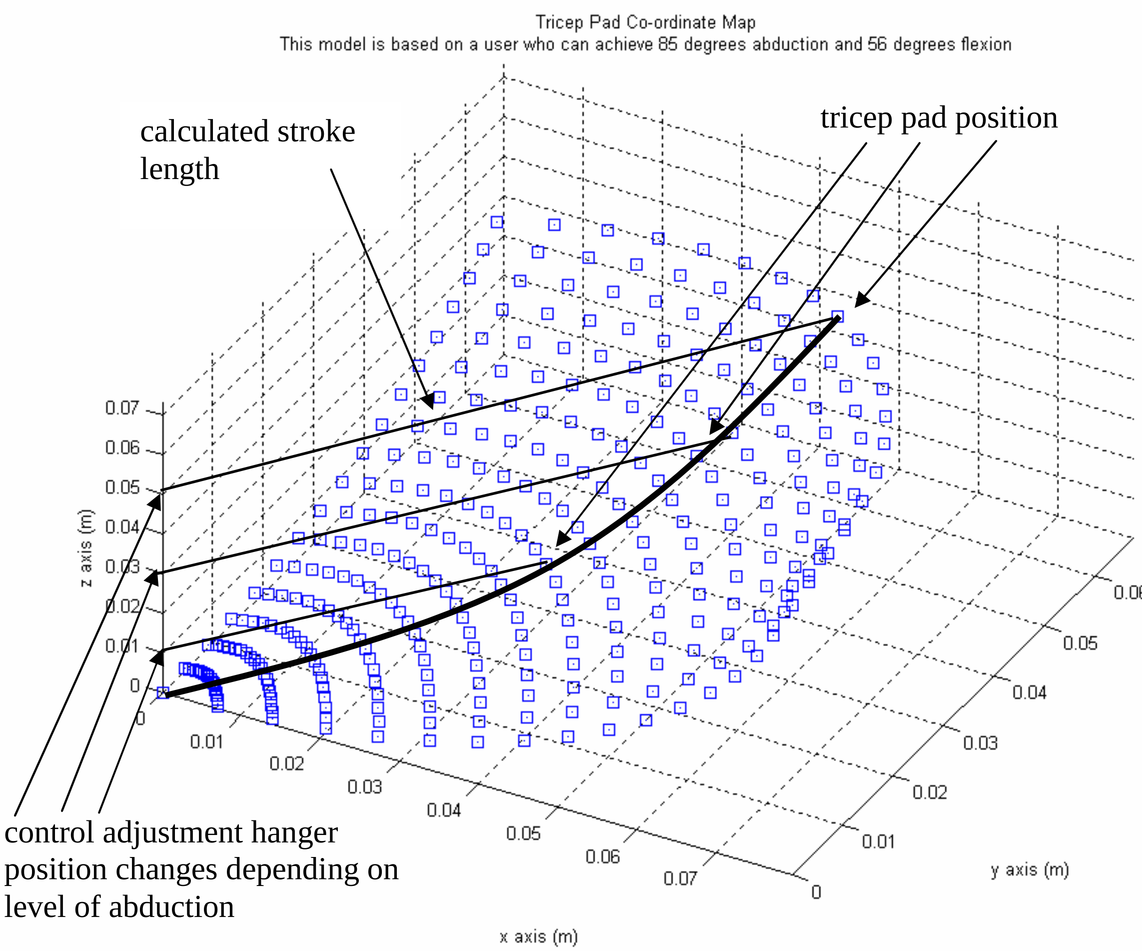


Figure 6-4 Alternative view of tricep pad positions with explanations of how the achievable stroke length was measured

Figure 6-5 shows the result for the calculated achievable stroke length based on the results of Participant #1. The figure shows the achievable stroke length on the y-axis, and the x-axis is the wrist rotational angle. This comparison was used because the orthosis was designed for the user's maximum achievable stroke to occur at -30° wrist extension, while the minimum achievable stroke length occurred at 30° extension.

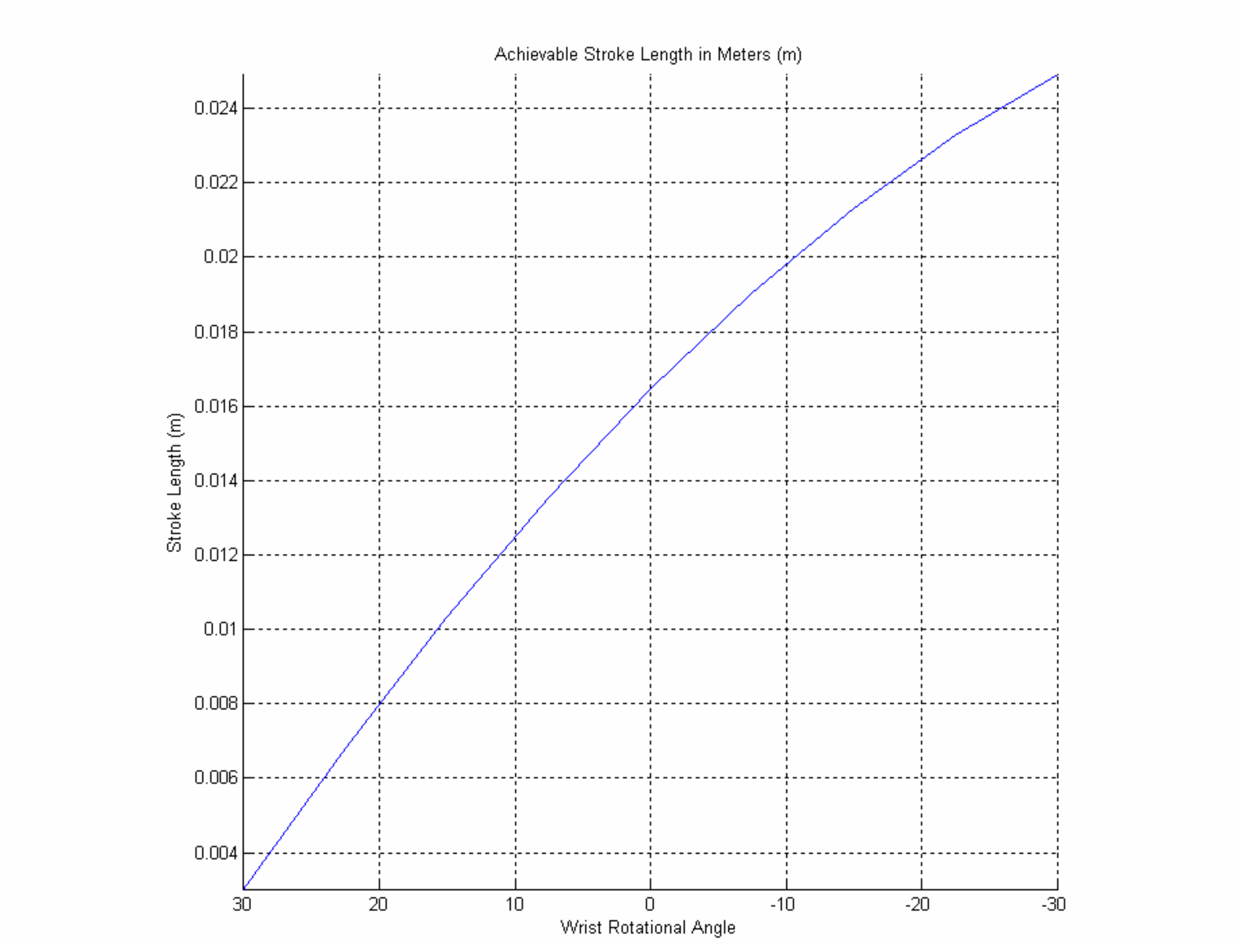


Figure 6-5 Achievable stroke length from the shoulder harness

The achievable stroke length as shown in Figure 6-5 contains one further change from the data shown in Figure 6-4. In Chapter 3, The Design Requirements Specification, Section 3.1.3 it was identified that there was a need for the user to freely move their shoulder around so that they could perform day-to-day activities such as picking up a pen, or drinking from a bottle. The *Functional Task Analysis* (Romilly et al., 1994) quantified these measurements over 22 activities and found that the user required a maximum 35° of shoulder elevation. The shoulder harness therefore was designed to allow the shoulder joint to move freely over the first 35° after which point the cable would tighten.

This was achieved by designing the harness such that the control adjustment strap was slack for the first 35° of shoulder elevation such that the user could move their arm around freely. After 35° of shoulder elevation the control adjustment strap would come into effect and start to pull on the cable. Therefore the achievable stroke length shown in Figure 6-5 has been modified to account for this.

6.1.2 *Achievable Force*

Shoulder abduction and flexion of the operational shoulder were the primary body motions used to power the orthosis, however the auxiliary shoulder protracted and retracted to give small, but important control to the harness.

The achievable forces were predicted based on the measurements from participant #1 (Table 4, P53). However, instead of using 36N in shoulder flexion and 28N in shoulder abduction, as was measured, the following force map used 30N in both abduction and flexion to simplify the model. The predicted achievable forces were based on the assumption that the forces linearly decreased from the maximum force, through to the minimum force of 0N. The *force map*, Figure 6-6, is the final result of the predicted shoulder strengths.

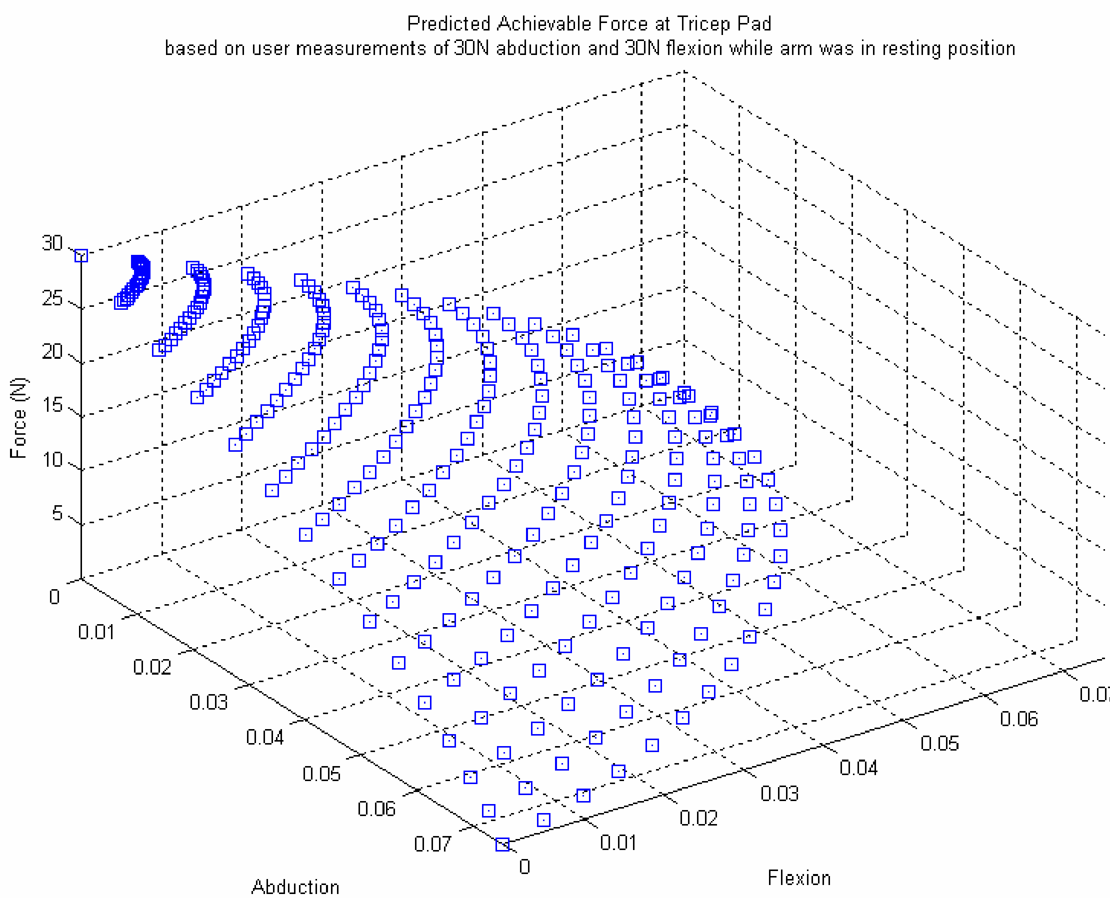


Figure 6-6 Force map from the shoulder

Figure 6-7 shows the achievable cable force from the shoulder. As identified in Chapter 3, The Design Requirements Specification, Section 3.1.3, it was important that the first 35° shoulder elevation was unutilized as it was necessary for use during activities of daily living. Hence, in Figure 6-7 the cable force starts at 14N and not at the maximum value of 30N. The force was compared to wrist rotation as it was designed such that the maximum force occurred at 30° extension and the minimum force occurred at 30° flexion (-30°).

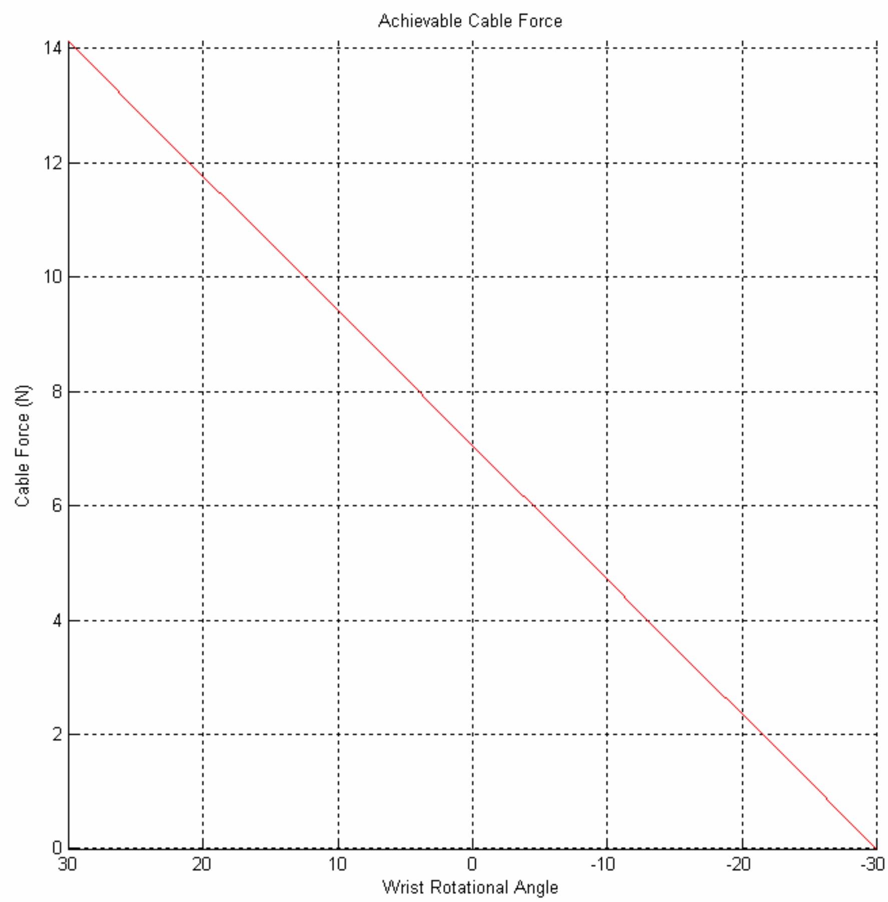


Figure 6-7 The achievable force generated from the shoulder

6.2 *Required Stroke*

The required stroke length was based on the kinematics necessary to pull the wrist from 30° through to -30° extension. Figure 6-8 shows the correlation of the physical model, to the Matlab model, which was used to calculate the required stroke length. The Matlab code used to create this model in shown in Appendix C, P117.

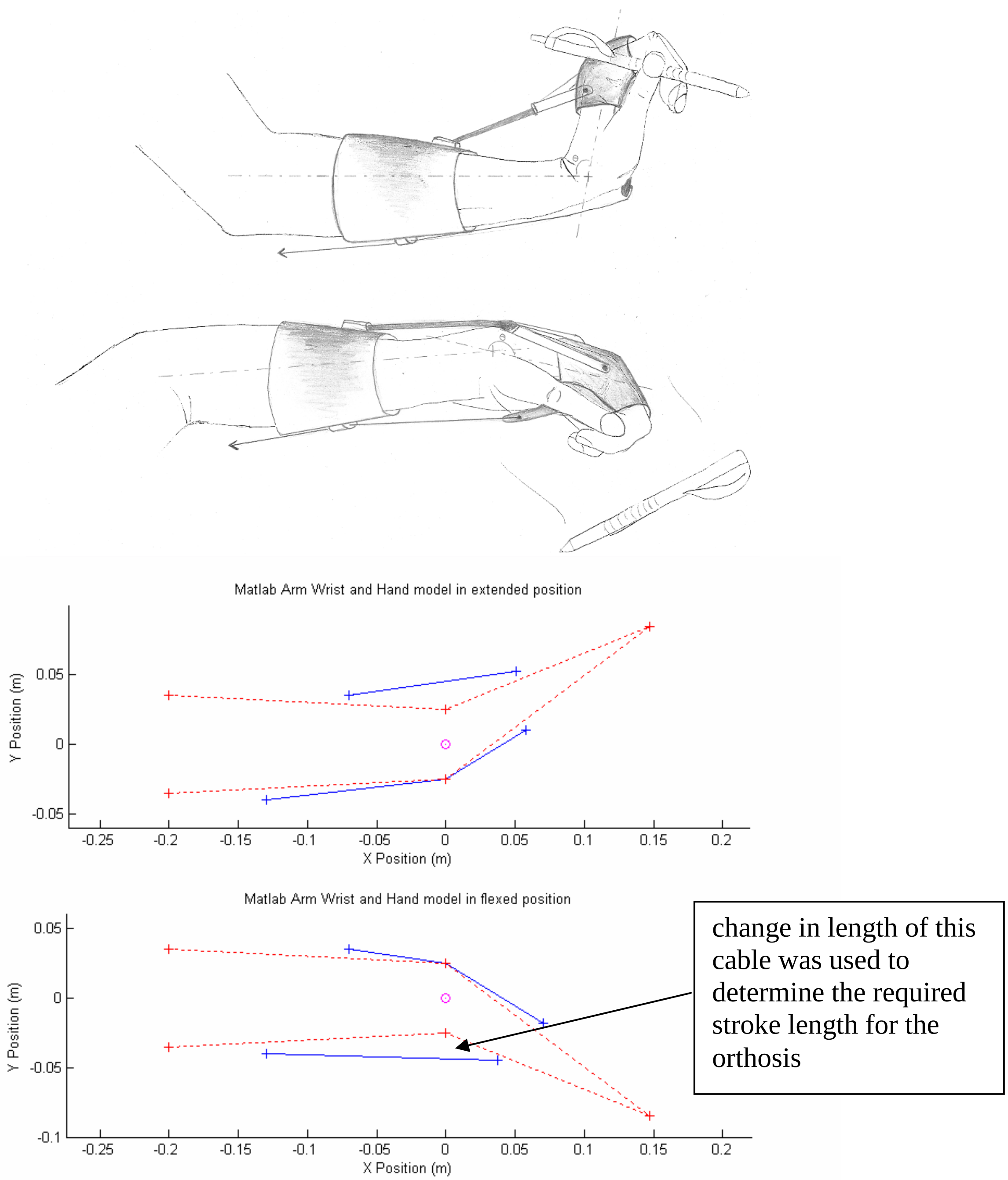


Figure 6-8 Comparison of physical and Matlab models

Figure 6-9 shows the theoretical required stroke.

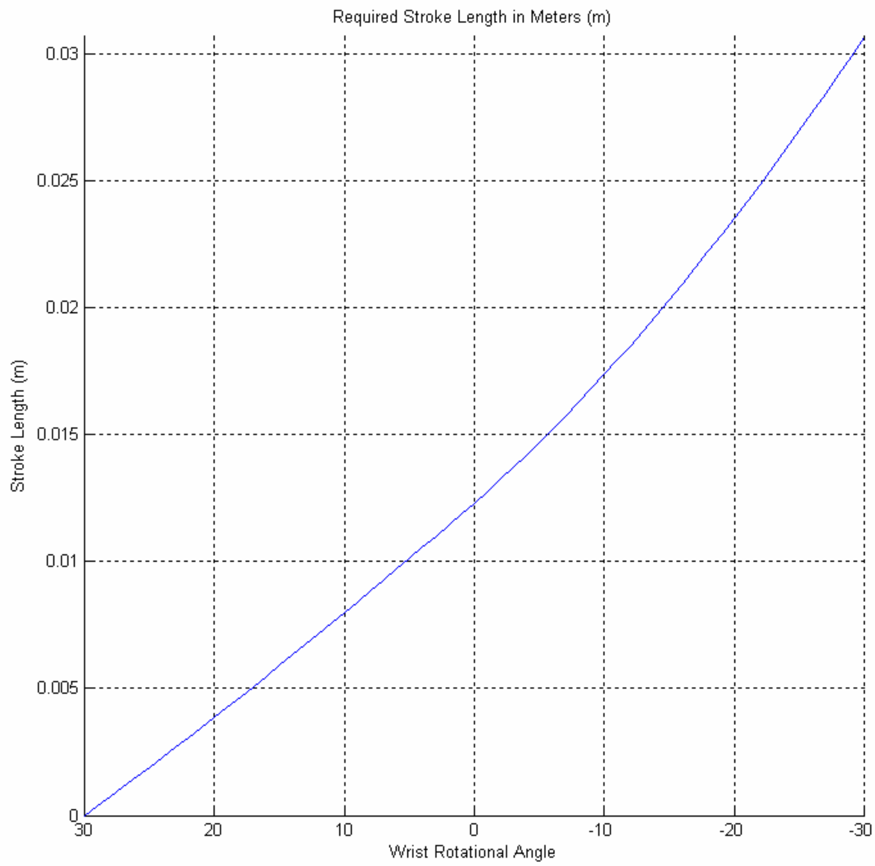


Figure 6-9 Required stroke length based on the Matlab model of the hand

Required Force

Figure 6-10 shows the wrist harness design for the hand, wrist and forearm. The horizontal hashed line running through the centre of the orthosis is the theoretical centre line of the forearm and hand. The wrist pivot position is the theoretical centre of rotation of the wrist, rotating about the head of the capitate bone (Jackson and Hefzy, 1994).

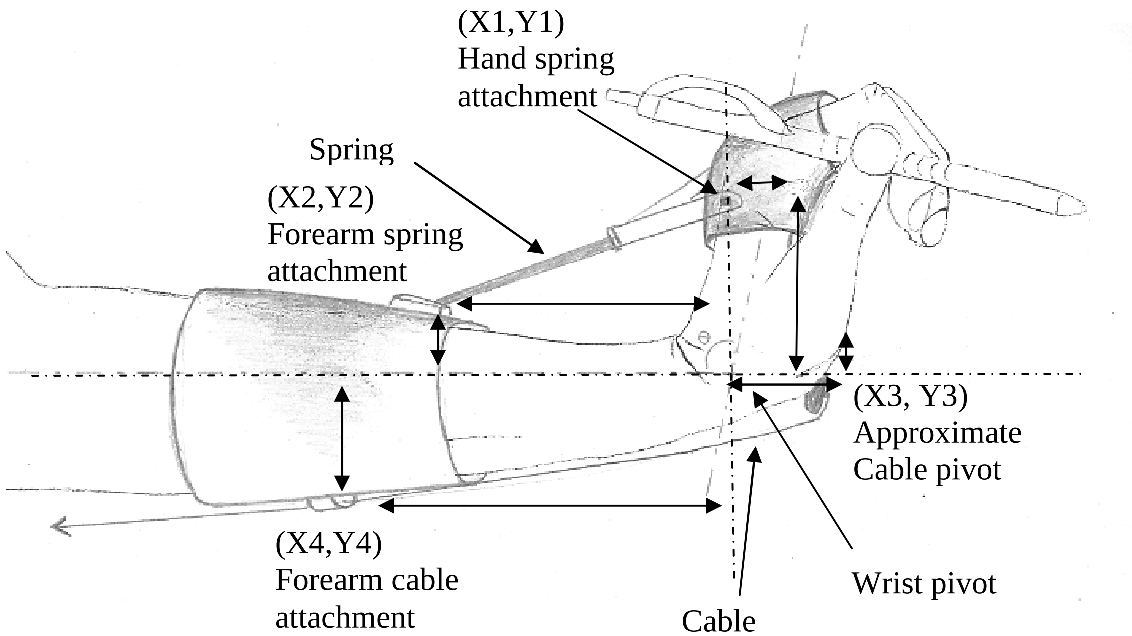


Figure 6-10 Wrist harness design

The required cable force was calculated based on the force required to pull the wrist from extension through to flexion. The spring, connecting between point X1,Y1 and X2,Y2 had a stiffness (k) of 900 N/m, this was based on early experimental results (Appendix D, P133). The centre of gravity of the hand was calculated based on the assumption that the mass of the hand approximated a triangular shape. From the Humanscale data sheet the length of the hand for a 50th percentile male was 21.1cm and weighed 600g. Therefore the centre of gravity was found to be 7cm based on the theory that the centroid for a triangle is 1/3 of its length (Shigley and Mischke, 2003).

For this analysis it was assumed that only a point contact was made between the spring and cable on the anterior and posterior surfaces of the wrist. In actual fact, the contact should have been approximated using a rolling joint with friction taken into account.

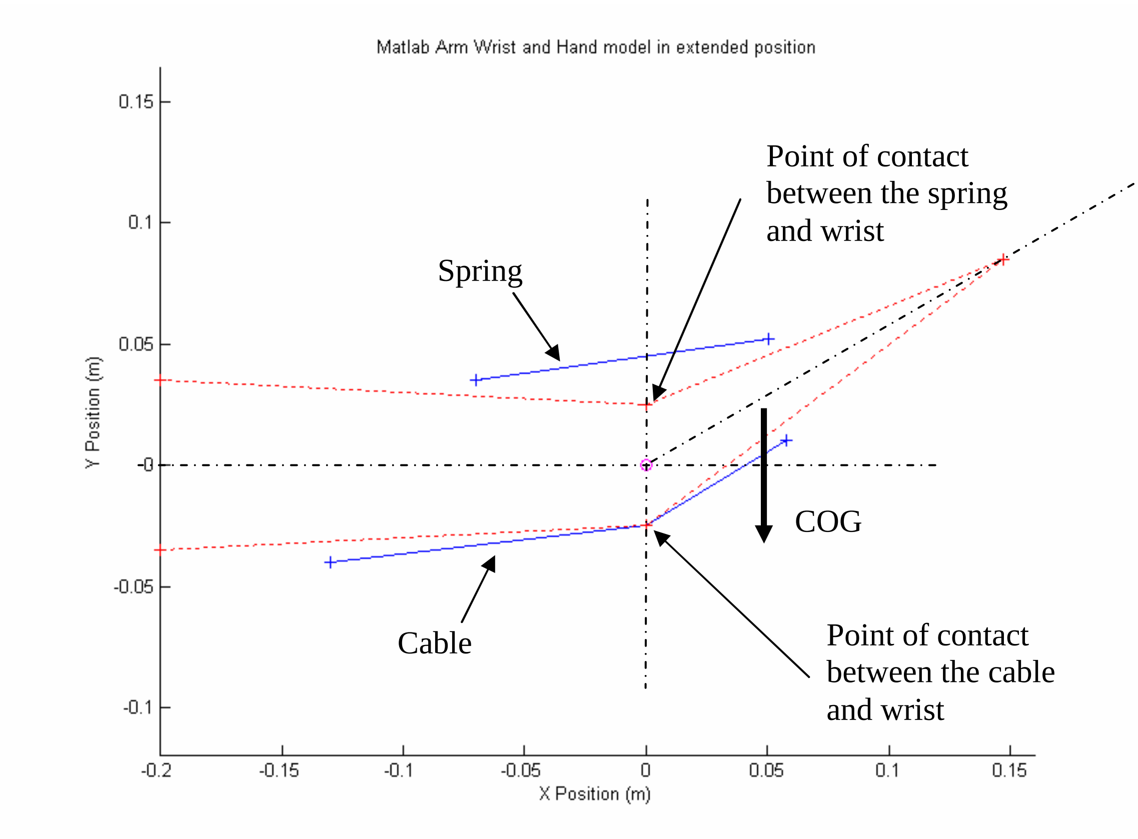


Figure 6-11 Overall wrist harness geometry

Figure 6-12 shows the required cable force to pull the wrist from extension through to flexion. The dynamics are a combination of the moment generated from the self weight of the hand, and the moment generated by the spring.

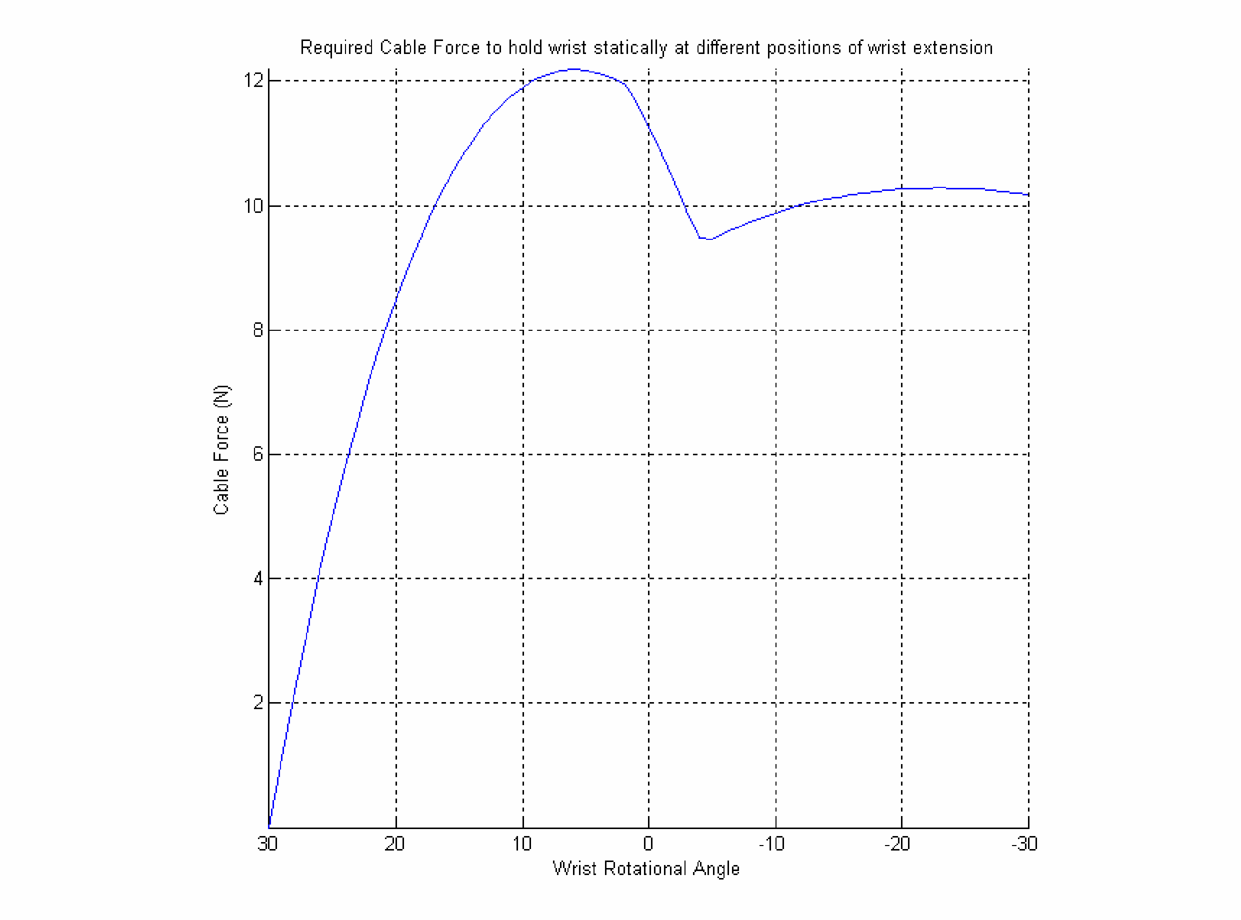


Figure 6-12 Required cable force to pull the hand from extension through to flexion

6.3 *Achievable vs. Required Energies*

The importance of analyzing both the achievable and required stroke as well as the achievable and required force was that it was possible to predict the functional outcome of the orthosis. For the orthosis to function correctly, it was necessary for the required force to remain below the achievable force. Figure 6-13 shows a breach at approximately 15° extension which indicates that the required force exceeds the achievable force. At this point the user would be unable to obtain any further wrist rotation.

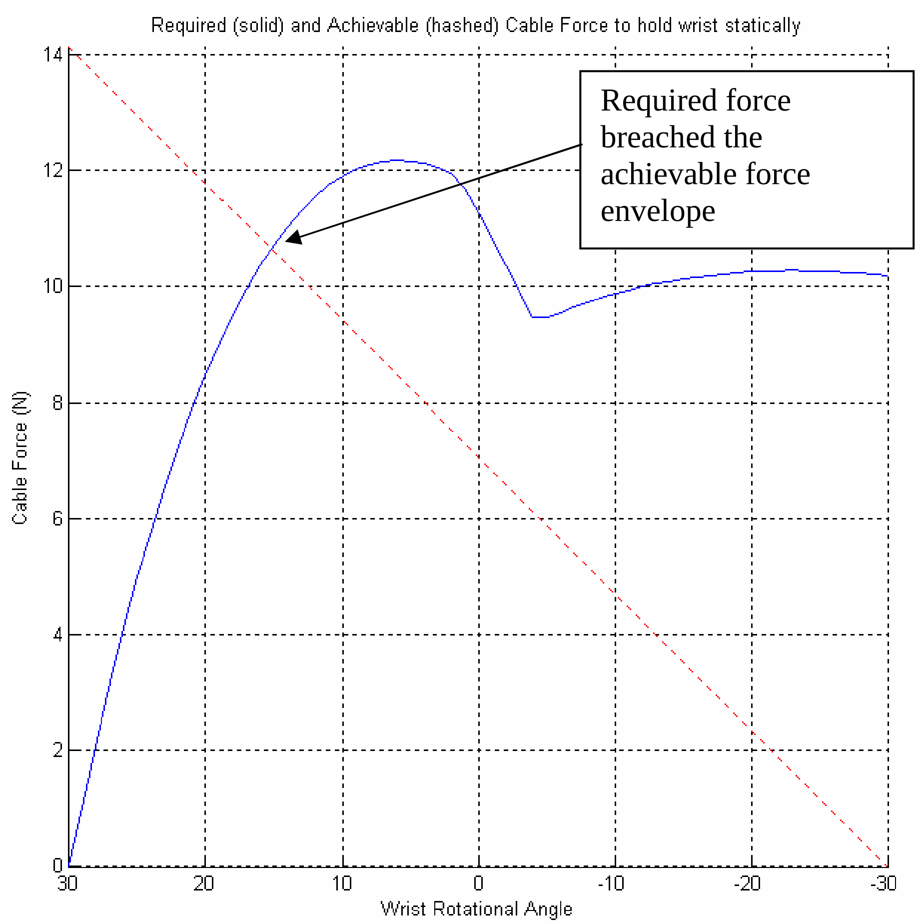


Figure 6-13 Required and achievable force comparisons

The second consideration was the required and achievable stroke. As before the aim was to identify the point where the required stroke exceeded the achievable stroke. Figure 6-14 shows the comparison of the required and achievable stroke lengths, and the breach occurring at approximately -18° extension. Because the required force exceeded the achievable force at 15°, this demonstrated that the force was the limiting factor for this orthosis design. However in some other cases the stroke may be the limiting factor as discussed later.

The aim of the stroke and force comparisons was to identify the attainable wrist rotation. Using this information it was possible to change the design of the orthosis to improve the attainable wrist rotation which improved the functionality of the orthosis.

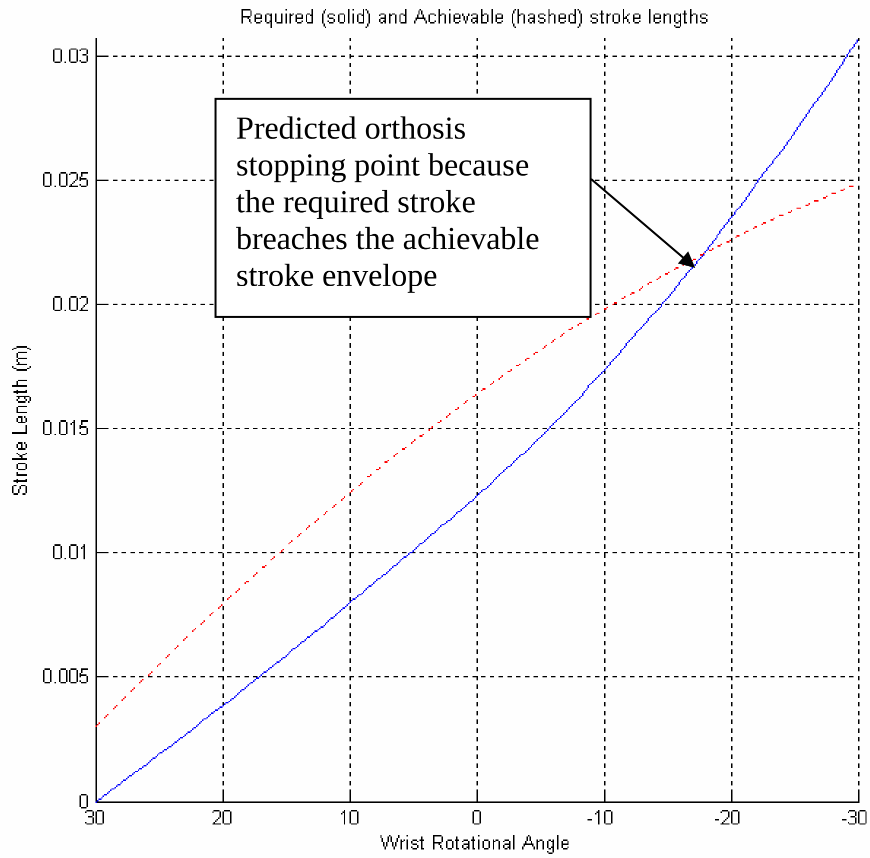


Figure 6-14 Required and achievable stroke comparisons

In summary, based on the theoretical model, the required force exceeded the achievable force at approximately 15° extension. This theoretical result suggested that the actual design would fall short of the target which was -30° extension. Because of this, it was necessary to improve the design by changing the geometric properties of the orthosis which is described in Chapter 7.

Chapter 7 Optimization

From the results of the force and stroke comparisons, it became clear that it was necessary to change the geometric parameters of the *self powered wrist extension orthosis* such that the required force was within the limits set by the achievable force, and the required stroke was within the limits of the achievable stroke. The aim was to change the shape of the required stroke and force graphs to minimize the attainable wrist rotation i.e. attain the greatest change in wrist rotational angle. Figure 7-1 shows an example of two variations for the required force; Variation 2 attained -8° wrist extension while Variation 1 only attained 15° extension. However, though Variation 2 attained a large change in wrist rotation, the consequence was that the required stroke increased. Therefore it was important to optimize both the force and stroke simultaneously to find the largest change in wrist rotation.

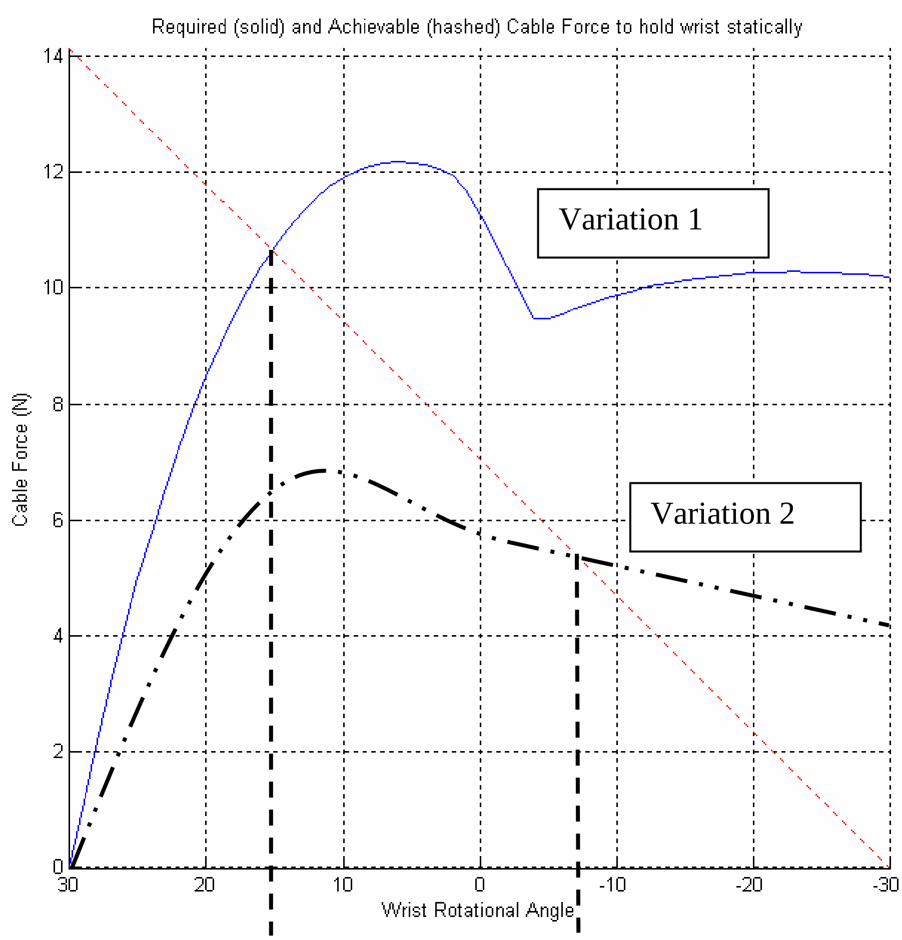


Figure 7-1 Example of improved orthosis design characteristics

7.1 Optimization Algorithm

The aim of the optimal design was to minimize the attainable wrist rotation angle by changing the geometric positions, X1,Y1; X2,Y2; X3,Y3 and X4,Y4 from the orthosis as shown in Figure 6-10, P62. Each one of these positions affected the required force and stroke, and because there were a large number of variables it was difficult to analyse each set individually. To resolve this, the Nelder-Mead unconstrained nonlinear search algorithm was used which was built into a Matlab function called, *fminsearch*. This search function was capable of handling large amounts of information as it searched for the local minimum of a specified parameter.

The algorithm was programmed to search for the attainable wrist position which was where the required force exceeded the achievable force, Figure 7-3. From this position the algorithm checked that the achievable stroke was greater then the required stroke. If the achievable stroke was greater then the required stroke then the original wrist position was past to the fminsearch function. However if the achievable stroke was less then the required stroke then the wrist angle was decreased to the point where the achievable stroke was equal to the required stroke as shown in Figure 7-4. The Matlab code for the optimization algorithm is shown in Appendix E, P135.

7.1.1 Example #1

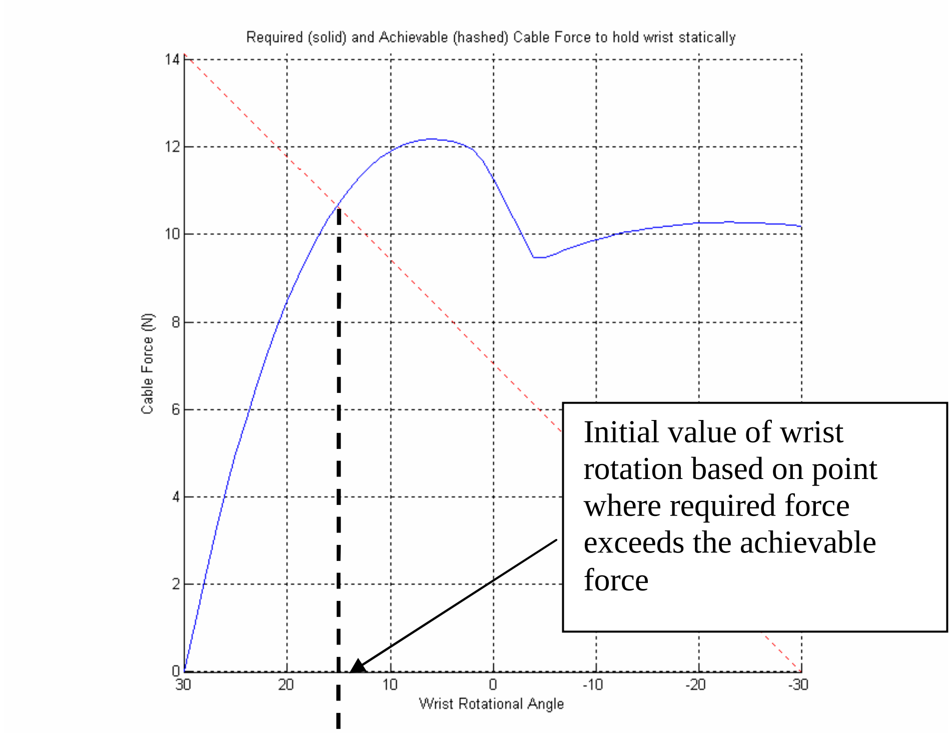


Figure 7-2 Identifies the positions where the required force exceeds the achievable force

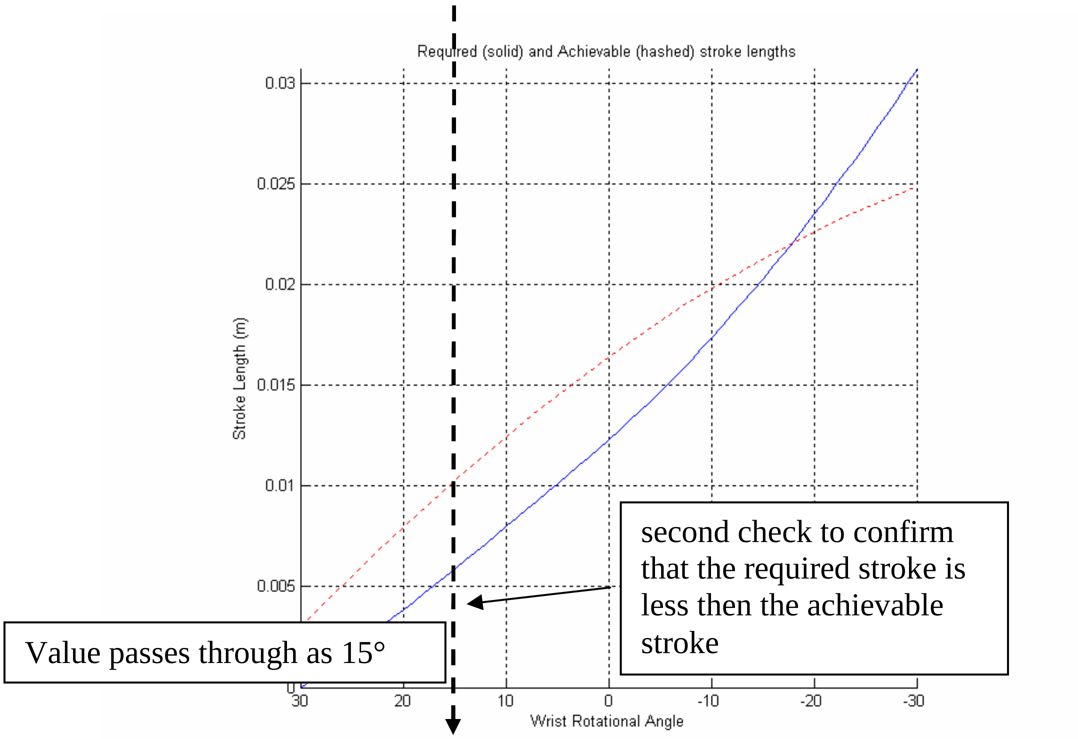


Figure 7-3 Check that the required stroke is less then the achievable stroke

7.1.2 Example #2

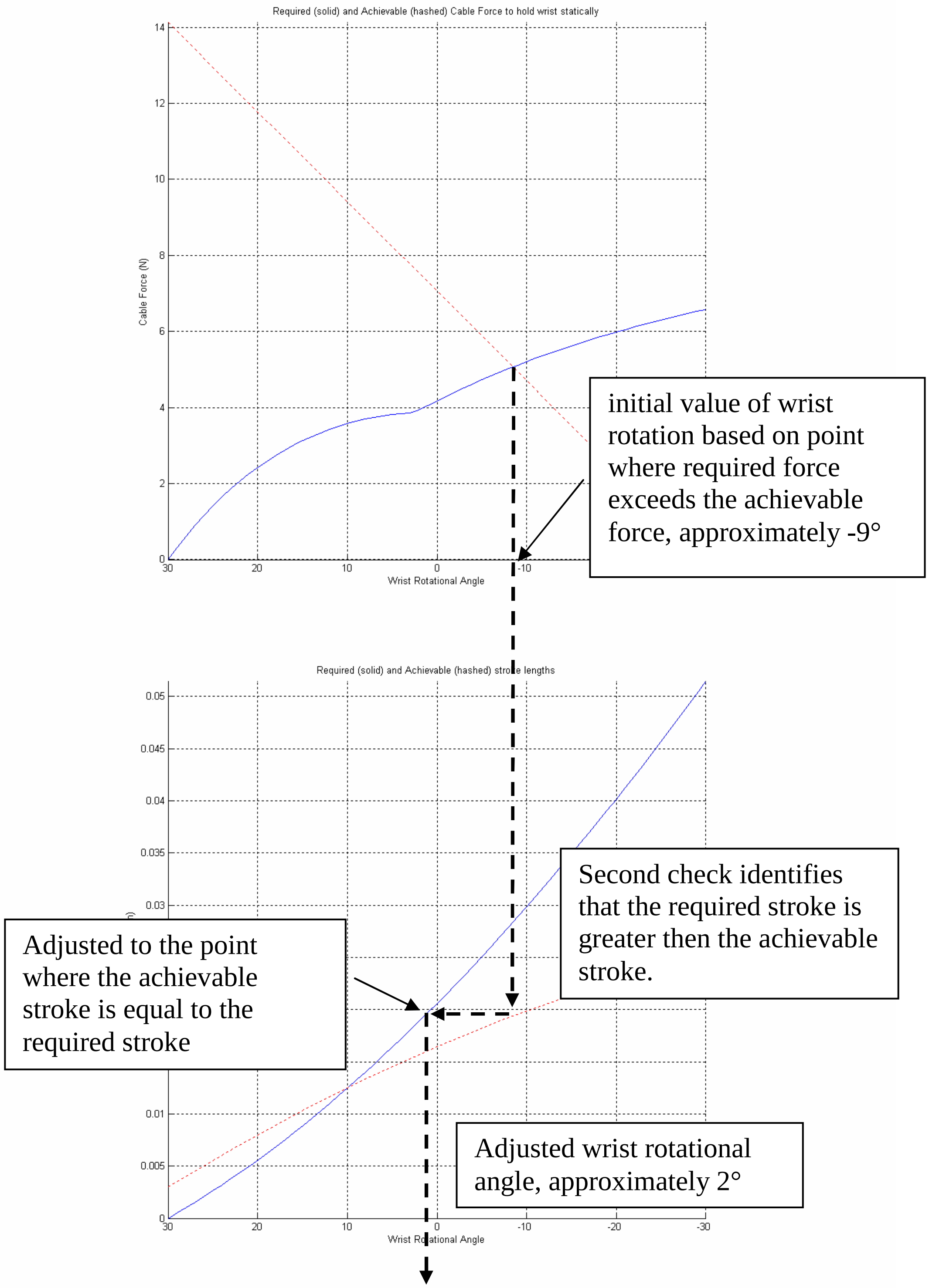


Figure 7-4 Attainable wrist angle algorithm part#2

The Matlab fminsearch function required an initial starting point. The initial geometric positions were chosen based on the fitting of the current concept design when worn by a male of the 50th percentile. Figure 7-5 shows how the values correspond to the orthosis, all values are in meters.

Table 5 Dimensions for the geometric positions of the orthosis

X1 = 0.07	Y1 = 0.02
X2 = -0.07	Y2 = 0.035
X3 = 0.055	Y3 = -0.02
X4 = -0.13	Y4= -0.04

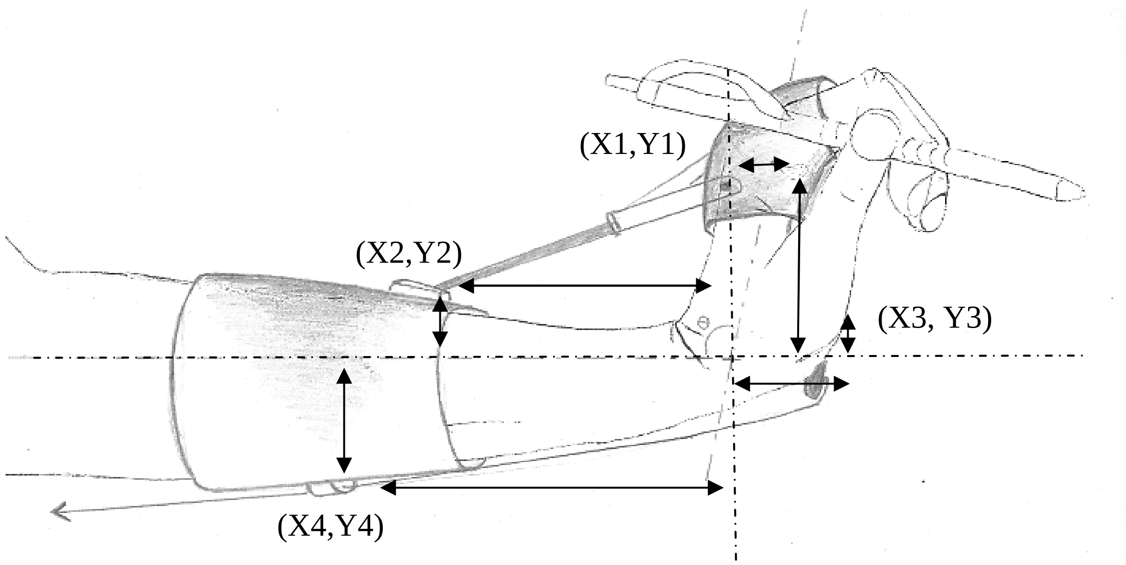


Figure 7-5 Wrist harness design

To ensure that the Matlab search function did not search for solutions outside of the desired range the following constraints were placed on the algorithm: The required force was always greater then zero to avoid instability; the geometric positions were restrained to the ranges shown in Table 6.

Table 6 Geometric constraints on the orthosis

$0.009 < X1 < 0.110$	$0.009 < Y1 < 0.110$
$-0.210 < X2 < -0.009$	$0.009 < Y2 < 0.110$
$0.090 < X3 < 0.110$	$-0.110 < Y3 < -0.009$
$-0.0210 < X4 < -0.009$	$-0.110 < Y4 < -0.009$

7.2 Optimization Results

The optimized design as found by the Matlab fminsearch function was as follows.

Table 7 Optimized results found using the Fminsearch function

X1 = 0.0652	Y1 = 0.0150
X2 = -0.0758	Y2 = 0.0366
X3 = 0.0606	Y3 = -0.0205
X4 = -0.1364	Y4= -0.0382

Figure 7-6 shows the current orthosis design, while Figure 7-7 shows the optimized design. It appears that there is very little difference between the two designs, the only significant feature is the slight change of the point X3,Y3 which has shifted slightly outward. The reason for the minimal changes in design was that Matlab search function located the local minimum rather then finding the global minimum.

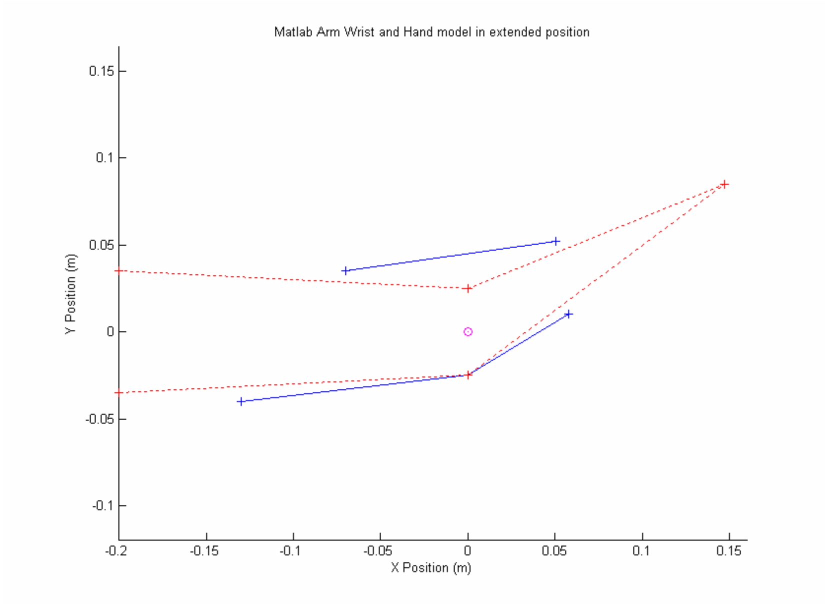


Figure 7-6 Current orthosis design

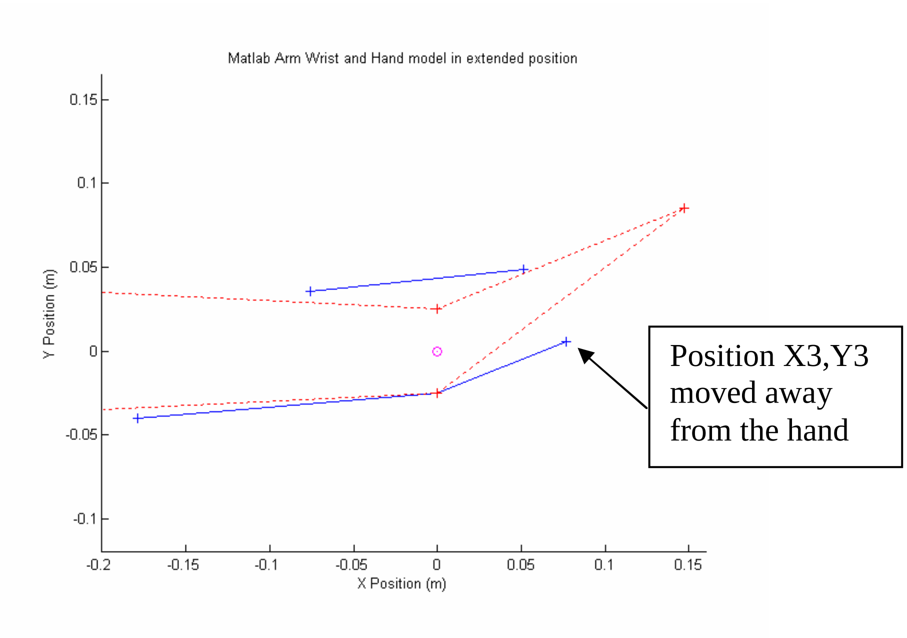


Figure 7-7 Optimized orthosis design

The results however for the wrist rotation were significantly improved as seen in Figure 7-8. The attainable wrist angle for the current concept was 15.2° while the attainable wrist angle for the optimized concept was -1.7°. This was a significant improvement between the two designs however it was still below the requirement of -30° extension.

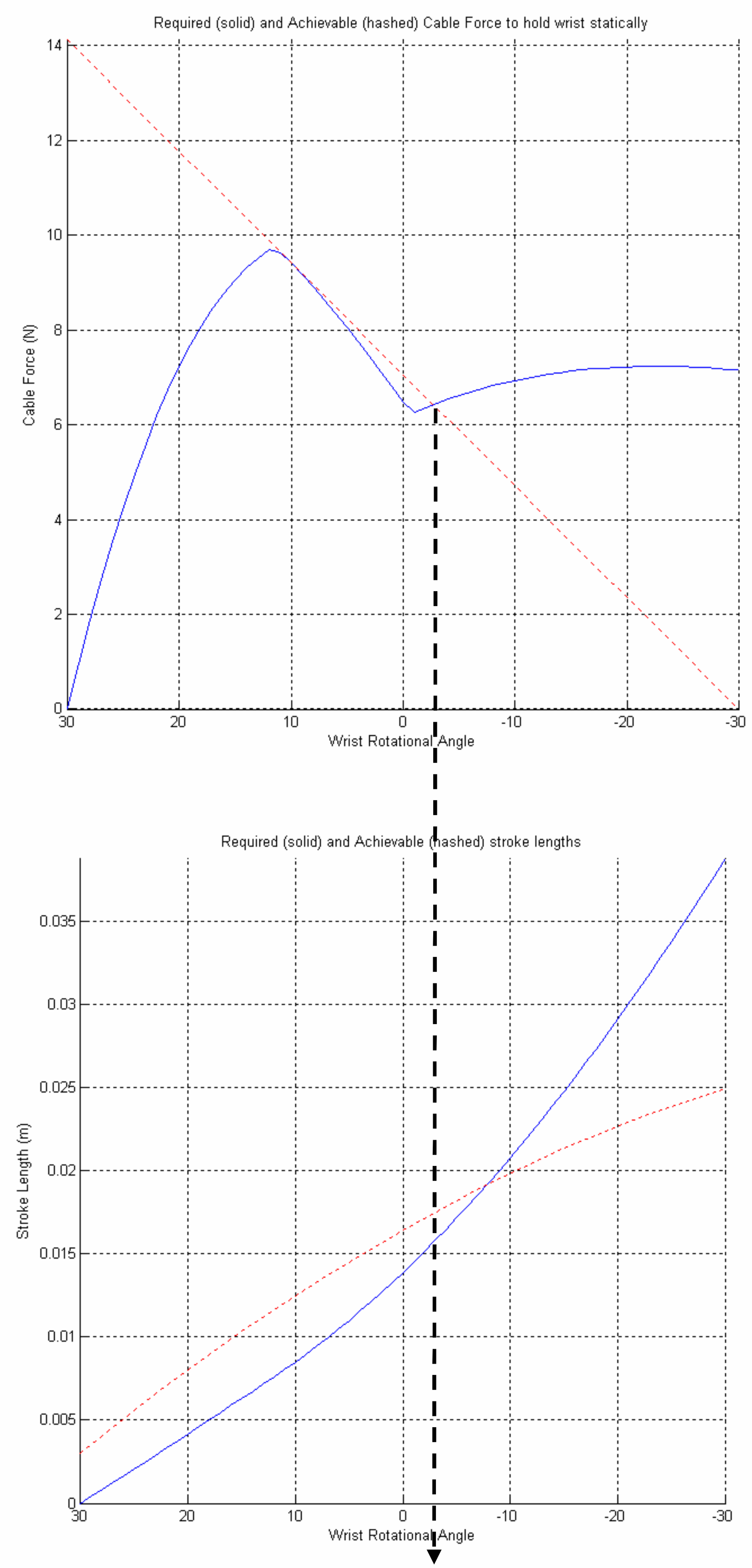


Figure 7-8 Achievable vs. required force and stroke characteristics for the optimized orthosis

7.3 Optimization Summary

The optimization results demonstrated that small geometric changes to the current orthosis design significantly improved the attainable wrist rotational angle. The geometric changes suggested by the optimization results showed that by changing point (X3,Y3) in Figure 7-7 away from the hand, reduced the required force to pull the wrist into flexion. Further analysis using a different optimization technique also concurred with this result.

The geometric positions of the current orthosis design were designed to minimize the visual appearance to improve social acceptance. As well as this the design requirements specifications identified the need to leave the palm of the hand uninterrupted to improve grasping functionality. Therefore the geometric positions were chosen to fit as close as possible to the skin to minimize these effects. The optimization results indicated an improvement by shifting the position X3,Y3 further out from the hand. However shifting this position would significantly reduce the aesthetics of the orthosis which is important as discussed in the design requirements. Even if the aesthetics were ignored the geometric changes suggested by the optimization results were so insignificant, that it was virtually impossible to create an orthosis that would be capable of adhering to the optimized specifications. The error involved each time the orthosis was put on by the user, would be significantly more then the changes suggested by the optimization results. Hence the result of the optimized orthosis was in actual fact more of a guideline rather then an actual specification.

Though the optimization process showed some significant improvement there were still a number of limitations of the model. Most significantly, the achievable force was predicted based on a fairly simplistic model of the shoulder strengths and data from only one candidate was used. Further limitations were that the required force did not account for friction in the wrist joint or friction generated between rubbing of the cable and skin. The model was also based on the assumption that the spring held the hand in extension against gravity. However during grasping activities the arm is often required to pronate or suppinate which would change the characteristics of the required force.

In summary, the optimization model predicted significant improvement with the attainable wrist position. However the geometric changes were very small that in actual fact they would be difficult to physically achieve. Combined with this the optimization results were limited because of the assumptions made for the achievable and required stroke calculations. Therefore the optimization result was recognized as a guideline to the orthosis design, but not necessarily the ‘optimum’ solution. The optimization process did not provide sufficient evidence to justify any changes to the current orthosis design.

Chapter 8 Final Concept

The user evaluation of the *self powered wrist extension orthosis* measured two outcomes which included: overall performance through visual inspections; and the attainable wrist rotation. The aim of the evaluation process was to determine if any discrepancies existed between the theoretical and experimental models. This was important because several assumptions were made in the theoretical model which may have not been appropriate or even necessary.

8.1 Final Concept Design

The *self powered wrist extension orthosis* is shown in Figure 8-1 through to Figure 8-4. The individual parts of the orthosis are shown in Appendix F, P147.

Figure 8-1 Front view of the orthosis

Figure 8-2 Behind view of the orthosis

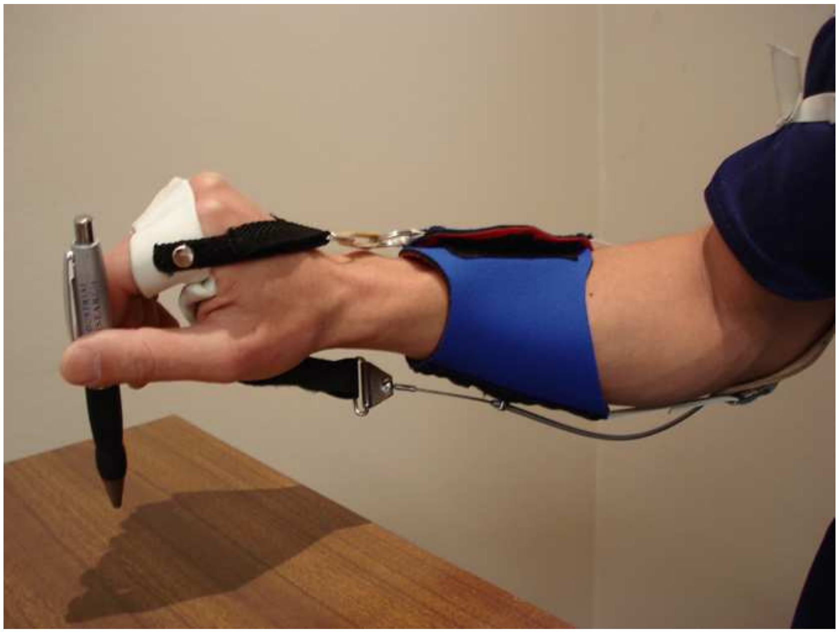


Figure 8-3 Arm Brace and Hand Brace during wrist extension

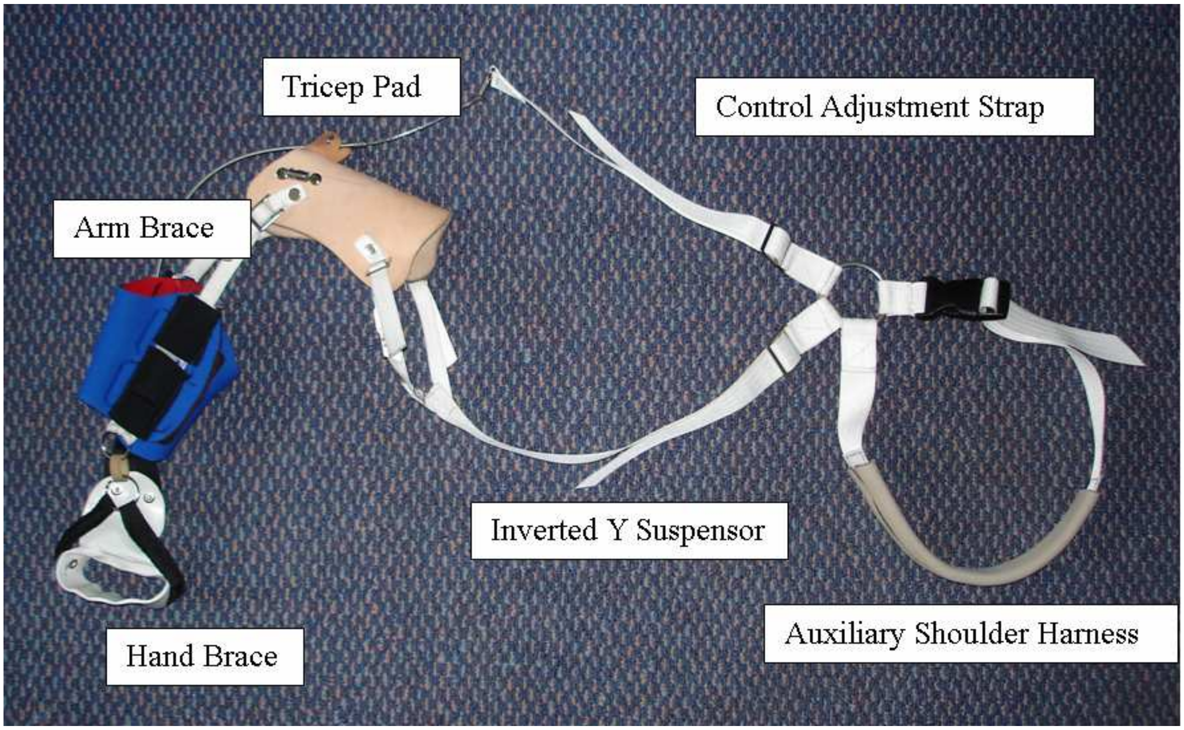


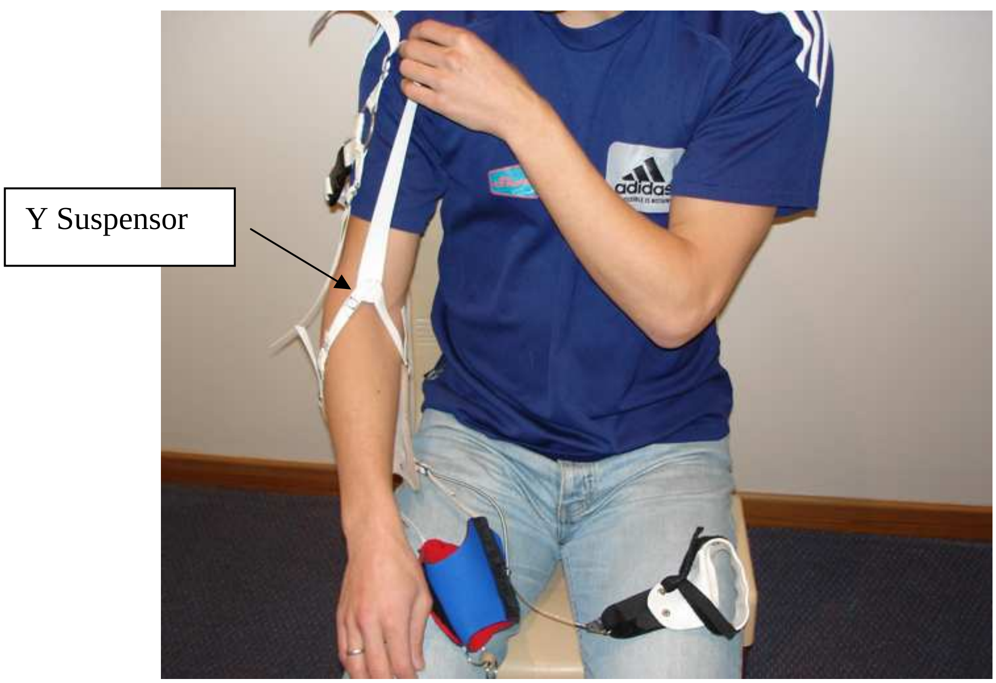
Figure 8-4 The orthosis laid out with explanations of the different components

8.2 Dressing

The design requirement specifications identified the need for unassisted fitting of the orthosis, however this was found to be very difficult to achieve. The solution to this requirement was a hybrid system where the user would receive assistance to fit the shoulder harness, however the hand brace could be fitted without assistance. The justification for this system was that during mornings and evenings a person with C5 tetraplegia receive assistance to help get washed and dressed. The hypothesis was that during the mornings the assistant would fit the shoulder harness and forearm brace. The user would then have the option to use the hand brace when they wished during the day. In the evening when the assistant returned, the shoulder harness and arm brace would be removed.

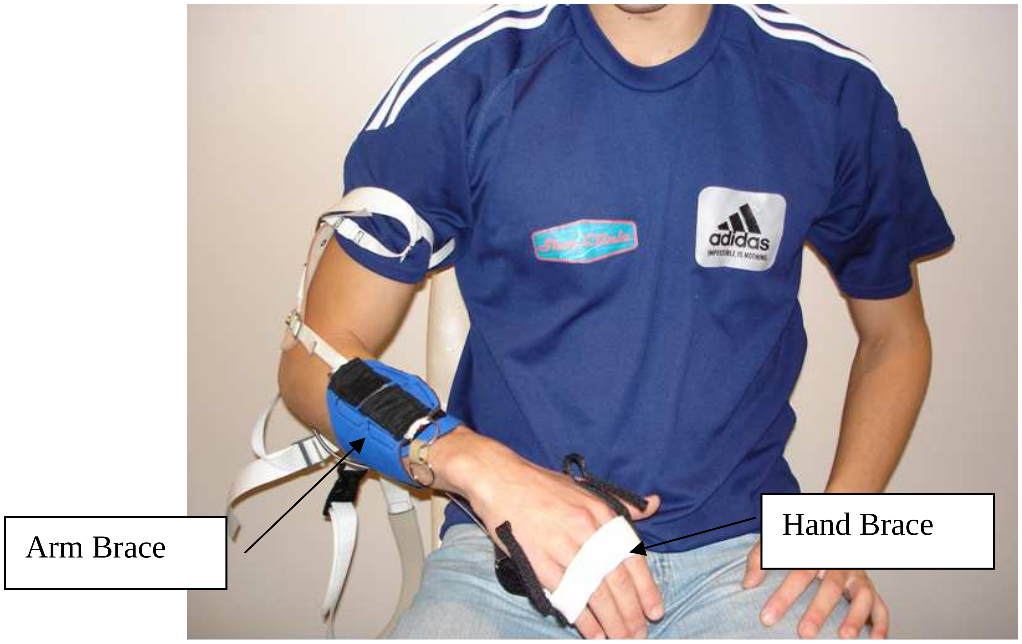
The current hand brace design as shown in the following figures requires assistance, however evidence suggests that self fitting is possible with some further minor design changes.

1. Slide hand and arm through Y-Suspensor



2. Attach arm brace

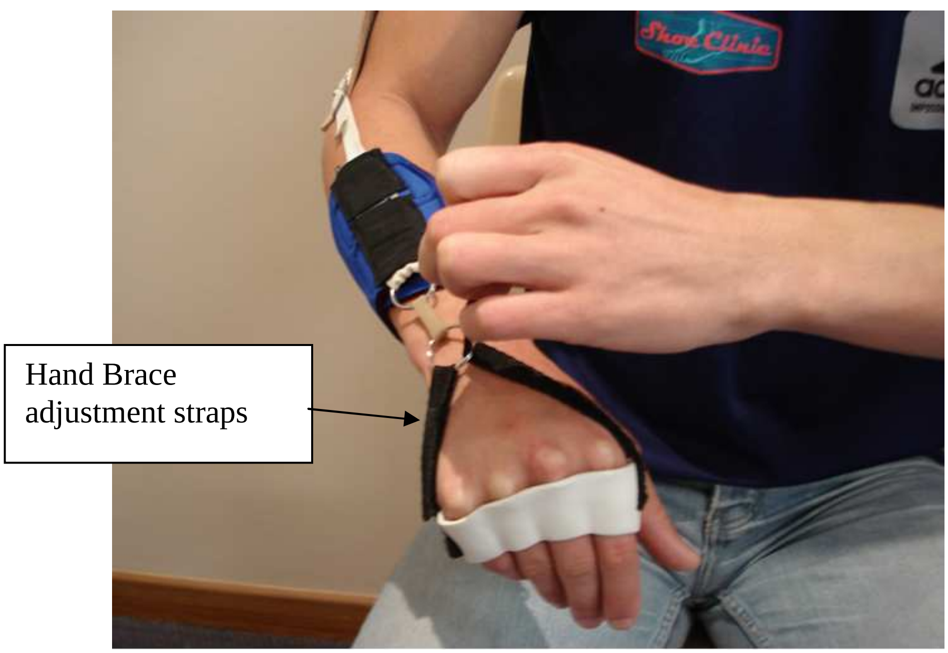
3. Slide hand through hand brace



4. Sling the shoulder harness across the back.

5. Attach buckle

6. Adjust hand brace appropriately



8.3 Using the Self Powered Wrist Extension Orthosis

To pick up an object the user must initially open the thumb, this is achieved by flexing the wrist forward which will open (abduct) the thumb. Wrist flexion is caused when the user reaches out for an object which increases the tension in the cable and pulls the wrist into flexion. The user is then required to place the object between the index finger and thumb.

To grasp the object the user is required to reduce the tension in the cable so that the wrist extends, this will pull the thumb towards the index finger and produce a ‘key pinch’ grip. To achieve this, the user must reduce the tension in the cable so the spring can pull the wrist into extension. Reducing the tension in the cable can be achieved through two means. Firstly by protracting and retracting the auxiliary shoulder which produces a small change in the cable length. However the primary shoulder motion is to decrease the elevation of the operational shoulder. This movement may change the position of the arm, hence, when the shoulder decreases elevation, the hand will be pulled away from the object. The solution to this problem is to exaggerate the elevation of the shoulder when reaching out for an object, thus the user can reduce elevation of the shoulder without changing the position of the hand.

The body movements described which are required to use the orthosis are slightly un-natural. However evidence suggests that given time, users of the device will learn how to perform these motions naturally. Figure 8-5 through to Figure 8-7 demonstrates how the orthosis interacts with the environment.

Figure 8-5 Reaching out towards an object

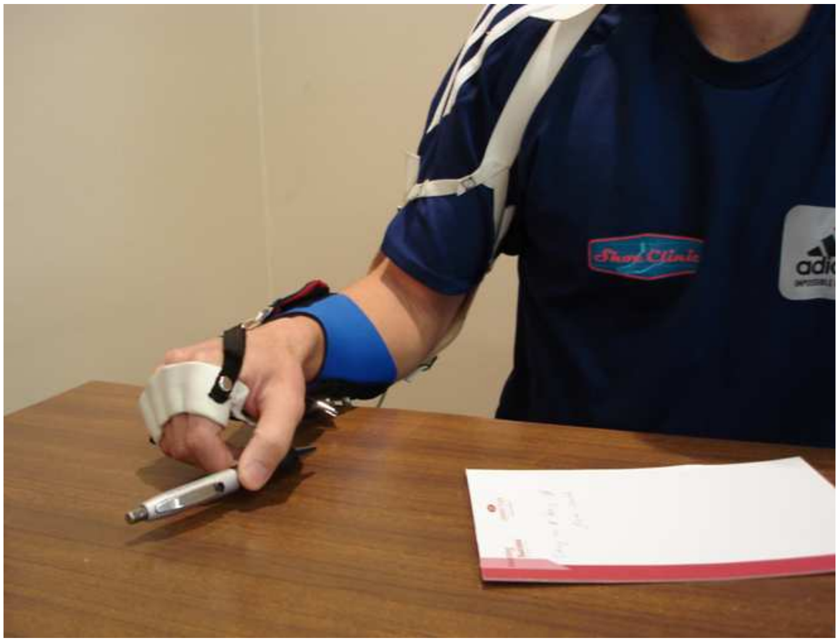


Figure 8-6 Grasping the object

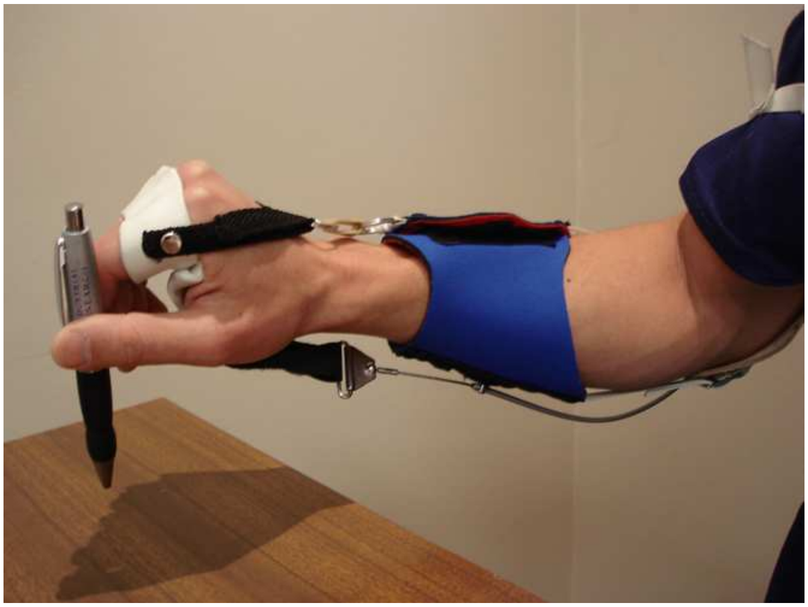


Figure 8-7 Elevation view of the orthosis holding a pen

Figure 8-8 shows the orthosis completely hidden except for the hand brace which can be easily put on, or taken off, by the user. Once the hand brace is removed, the shoulder harness automatically relaxes and leaves the user free of any restrictions.

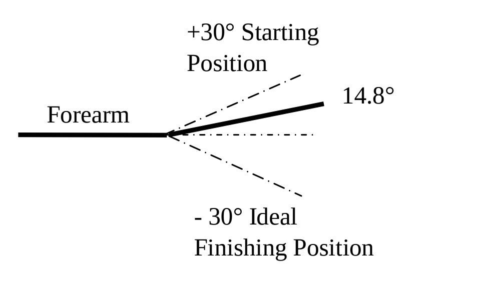
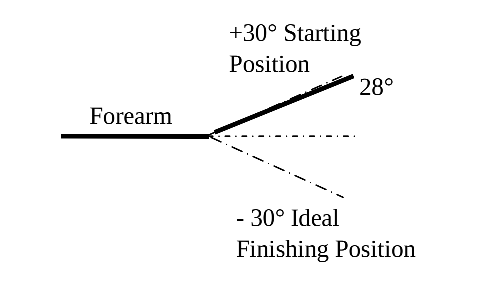
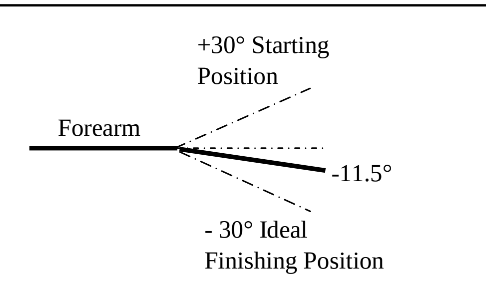
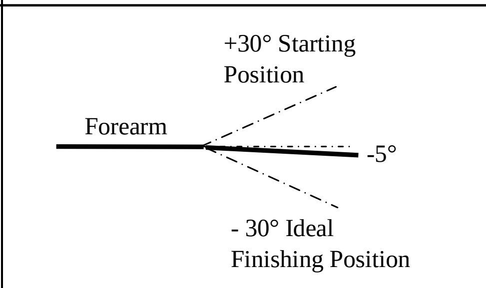
Figure 8-8 The orthosis is hidden under clothing, also a demonstration of holding different sized objects

Chapter 9 Results

9.1 Experimental Results

The *self powered wrist extension orthosis* was trialed on the same two participants whose shoulder measurements were used in the theoretical model. Table 8 shows the comparison of the theoretical and experimental results, with Participant #1 (the weaker of the two) attaining 28° extension, and a theoretical result prediction of 14.8°, and Participant #2 attaining -5° extension, with a prediction of -11.5°.

Table 8 Theoretical versus experimental orthosis results

	Theoretical Result	Experimental Result
Participant #1		
Participant #2		

9.2 Results Discussion

9.2.1 Discrepencies between theoretical and experimental results

1. In the theoretical model it was assumed that the support strap would have no effect on the dynamics of the orthosis; however the experimental trials demonstrated that the support strap did in fact have a significant impact as when the shoulder was elevated, the tension in the support strap increased. Because the support strap is inelastic, it pulled on the tricep pad and in turn the arm brace, which increased the extension of the spring as shown in Figure 9-2. The result was that the user was required to provide a greater force in the cable to flex the wrist, but because the user lacked strength to do this, the orthosis underperformed.

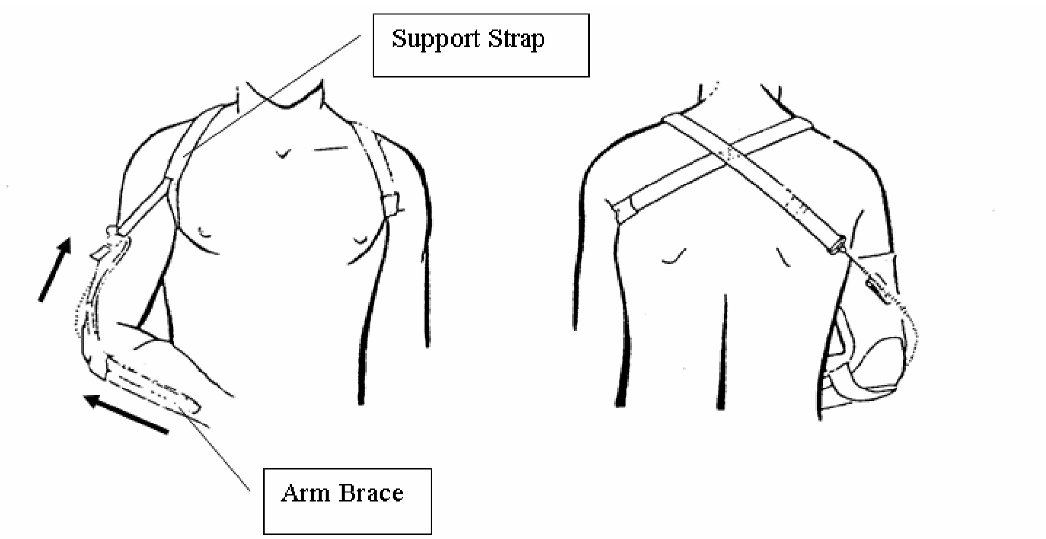


Figure 9-1 Characteristics of the support strap

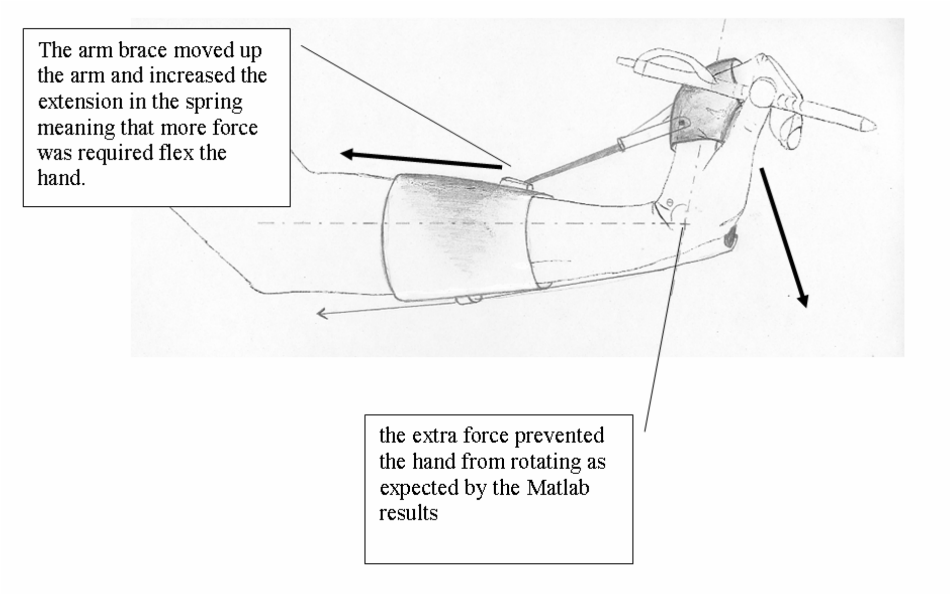


Figure 9-2 Increased force in the spring

2. During the experimental trials, friction was apparent between: (1) the control adjustment strap rubbing on the back of the body; (2) the inner and outer Bowden Cable rubbing against each other, and (3) and the internal losses within the body systems i.e. the wrist joint not acting as a perfect hinge. Friction has not been quanitified in this research, however based on visual approximations it is estimated that the orthosis performs within 70- 80% efficiency.

9.2.2 Design Improvements

1. The current wrist harness was restricted by the constraints on the geometries which were governed by the dimensions of the hand and arm. This significantly limited the possible positions of the spring attachments and cable attachments, making it impossible to define more desirable force and stroke characteristics.

2. The shoulder strength used in the theoretical model was based on the maximum static achievable force. However, during the experimental trials it became apparent that the user could not generate this force repeatedly; hence making this measurement inappropriate for the model. Instead the dynamic forces of the shoulder should be used and rather than measuring the maximum dynamic forces, the user should work in a range which feels ‘comfortable’ on a day-to-day basis.

9.3 Design Performance

The performance of the orthosis was evaluated against the criteria outlined in the design requirement specifications as shown below in Table 9. All design requirements 'passed' the design performance, apart from three whose results were 'unknown', and two who 'failed'. It was unfortunate that the two requirements that 'failed' were the most important to the solution success, with (1) being the 'achieved range of motion of the wrist' which was required to extend and flex to 30°, and (2) the 'speed of rotation'.

Table 9 Design Performance of the Orthosis

DESIGN PERFORMANCE			
FACTORS	**CONSIDERATIONS**	**OUTCOME**	**COMMENT**
BODY FUNCTIONS	Maximum weight 200grams on forearm and wrist	Passed	113grams and total weight of 283grams
	Provide 30 degrees wrist flexion	Failed	Attained only 28°
	Provide 30 degrees wrist extension	Passed	
	10N Pinch Force	Unknown	
	1Nm - 2Nm of wrist torque	Passed	The wrist was held firmly in extension suggesting that this result was achieved
	Operate at 60 RPM	Failed	Friction resulted in low performance
	Free movement of the thumb	Passed	Hand Brace design allowed for movement
	Fixed finger position to provide lateral support for the key pinch	Passed	Hand Brace design provided platform for key pinch through controlling finger position
	Controllable	Passed	Achieved through Bowden cabling system
	Accurate Control of wrist position in both Extension and Flexion	Passed	Achieved through Bowden cabling system
	Palm of hand left un-obstructed	Partial Pass	Though not completely un-obstructed the Hand Brace did not affect grasping activities
ACTIVITIES			
	First 35° of shoulder elevation remains unrestricte d	Passed	Achieved through slack in Control Cable
	Auxiliary hand remains free of restrictions	Passed	
	Natural wrist motion	Passed	No mechanical coupling of wrist hence the wrist moved with natural kinematics
	Adduction and Abduction wrist control	Passed	Hand Brace provided control of both abduction and adduction
	Mobile on wheelchair	Passed	
	Mobile on body	Passed	
	Responsive	Partial Pass	Due to the large required force this was difficult to evaluate
PARTICIPATION			
	Safe	Passed	
	Quiet	Passed	Virtually no audible sound
	Hidden under clothing	Passed	
	Unassisted fitting and unfitting	Partial Pass	Shoulder harness required assistance however the wrist harness did not
	Soft materials	Passed	
	Breathable	Passed	
	Durable	Passed	
	Robust and Waterproof	Passed	
	1 min fitting time on and off user	Passed	Approximately 1min was necessary found during user trials
ENVIRONMENTAL			
	One size fits all	Passed	
	Common engineering materials	Passed	
	Provide training	Unknown	
	Superior to competition devices	Unknown	
PERSONAL			
	Simple techonology	Passed	
	Short training time 1-2days	Passed	An expected result

Chapter 10 Further Development

An *alternative wrist harness* design was developed as shown in Figure 10-1 and Figure 10-2, which utilizes the same shoulder harness as the current design. The advantages of the alternative design included greater versatility of the required force characteristics, improved aesthetics and functionality (the cable does not run underneath the arm and hand), and eliminating the spring from over extending as it is connected by a common base.

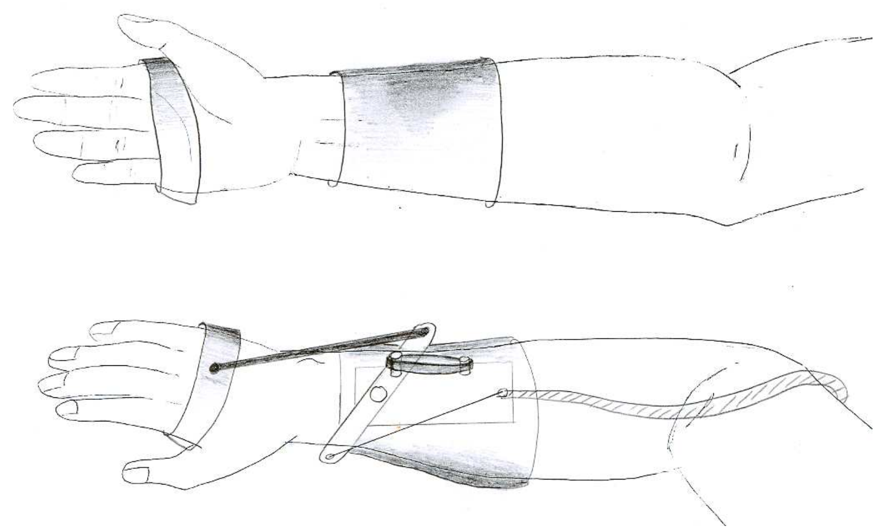


Figure 10-1 Alternative hand mechanism design, showing the wrist in extension

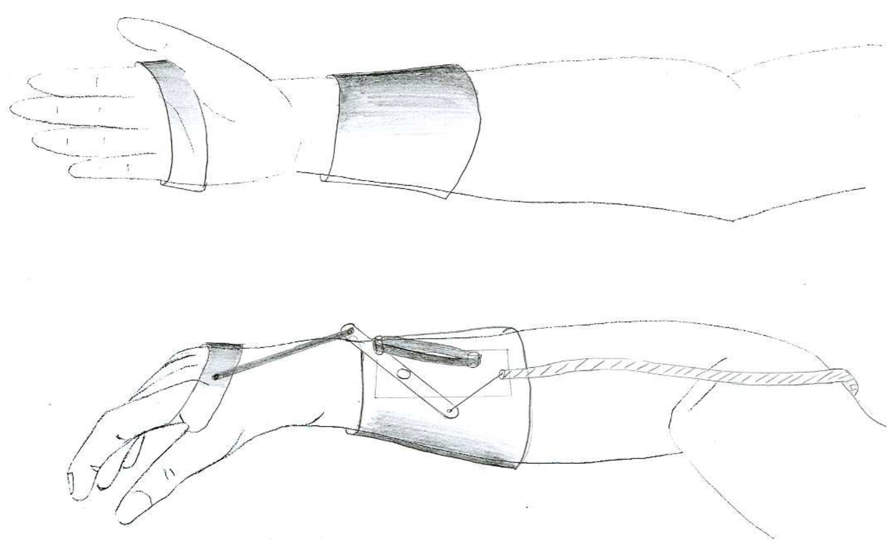


Figure 10-2 Alternative hand mechanism design, showing the wrist in flexion

The Matlab results for the alternative design demonstrated many desirable characteristics. Figure 10-3 shows a schematic view of the alternative wrist harness, with the diagrams on the left column being the plan view, and the columns on the right the side view.

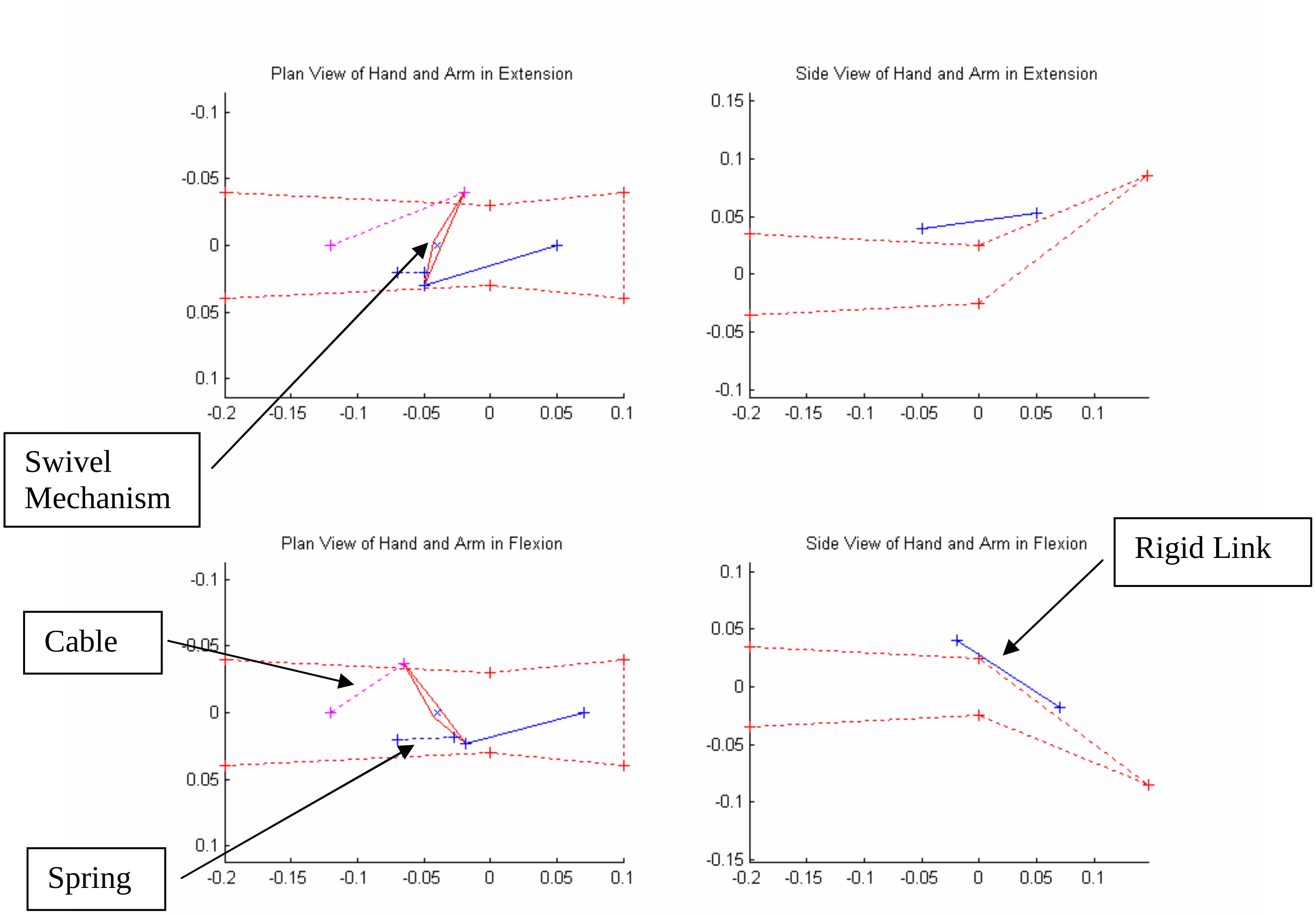


Figure 10-3 Theoretical model of the alternative design. The red hatched line is the profile of the hand and wrist. The upper two diagrams show the hand in extension, while the bottom two show the hand in flexion

The *alternative wrist harness* functions exactly the same way as the current wrist harness where the user tightens the cable by manipulating the shoulder harness, turning the swivel mechanism and pushing the hand into flexion. The major change is a rigid link which connects the mechanism to the hand brace as it is required to support compression.

The theoretical results for the *alternative wrist harness* is shown in Figure 10-4 and Figure 10-5 and the Matlab code are shown in Appendix G, P153. These results were encouraging as the characteristics of the required force were relatively low meaning that a greater attainable wrist rotation was achieved. The required stroke also reduced, suggesting that this alternative design had superior characteristics than the current *wrist extension orthosis* design. The attainable wrist position for Participant #1 was approximately -25°.

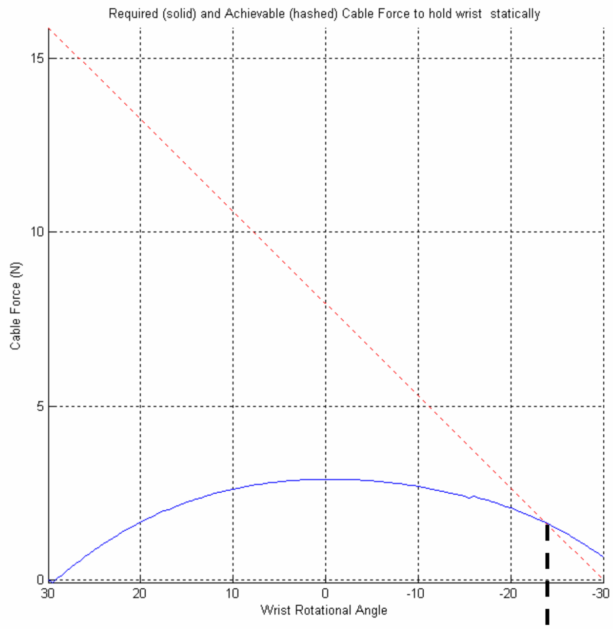


Figure 10-4 Required vs. Achievable force for the alternative design

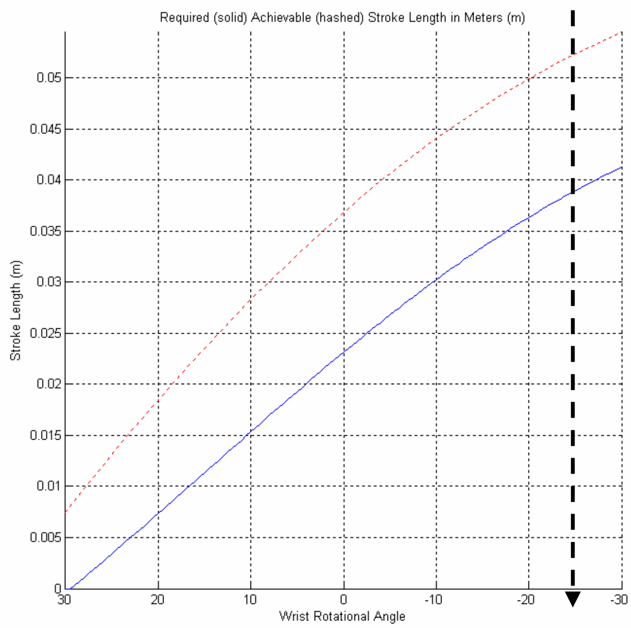


Figure 10-5 Required Vs Achievable stroke for the alternative design

Chapter 11 Summary

It is estimated that the annual incidence of spinal cord injury (SCI), not including those who die at the scene of the accident, is approximately 40 cases per million. Out of the various levels of spinal cord injury, the most common level of impairment was C5 tetraplegia at 14.8%. The disabilities associated with C5 tetraplegia are complete loss of function to the lower extremities including the abdomen and bowel. The shoulder and arms are still functional, however wrists and hands are completely paralyzed making simple tasks such as holding a pen or eating with a fork virtually impossible. Fortunately there are some simple solutions to overcome this. For example, hand splints have a slot where a spoon can be inserted, and because the wrist is statically braced, the users are able to feed themselves. However, this solution has limitations as it does not help the user to grasp a napkin or perform other simple tasks. The solution to these problems is to provide the affected individuals with a means to grasp and manipulate objects with the use of a ‘key pinch’ grip.

Surgery is generally accepted as the standard method to restore a key pinch grip for someone with C5 tetraplegia, where the thumb closes on the lateral side of the index finger To achieve this through surgery, wrist extension is required to provide control of the wrist followed by an operation to the thumb, so that when the wrist extends, the thumb forms a key pinch grip. Generally, wrist extension is achieved by taking an active muscle and transferring it onto a wrist extensor tendon. In most cases this operation is successful, but in other cases, a person with C5 tetraplegia may not have control of the active muscles which are necessary to perform this operation. Even if there is an active muscle with sufficient strength, the surgery is very complex and sometimes has poor results.

An alternative to surgery is assistance using a mechanical solution. For many years engineers and surgeons have been developing solutions to provide a person with C5 tetraplegia with the means to grasp and manipulate objects. The current devices in the market targeted to help those with C5 tetraplegia include the Freehand System, The NESS Handmaster, and the Power Grip. However, it seems that while one device will work for a certain group of individuals the same device will not work for another and discontinuence of products seem to be common practice.

The design requirements for the *self powered wrist extension orthosis* were determined from the International Classification of Disability and Health (ICF), which is used a standard measurement tool to represent a person's 'quality of life'. Findings from the ICF suggested that an acceptable device required not only functionally, but also practicality and social acceptance. The ICF also demonstrated the importance of Personal and Environmental factors which included ceratin aspects such as training, cost, assistance needs, gender, and age differences.

The *self powered wrist extension orthosis (orthosis)* was created using a systematic design process based on the design requirement specifications. The orthosis was seperated into two sub-sections which included the shoulder and wrist harness. The shoulder harness was re-modeled from the currently used prosthetic limb design that contains a brace which stretches across the users shoulders and extends a cable and curtails as the user manipulates their shoulders. The wrist harness uses this cable to pull the wrist into flexion while a spring counters this force and holds the wrist in extension. Combining the orthosis with a tenodesis surgery of the *flexor polices longus* provided a 'key pinch' grip secondary to wrist extension. The operation consisted of anchoring the *flexor polices longus tendon* to the bone, hence as the user extended their wrist, the thumb would 'pull' towards their lateral side of the index finger forming a 'key pinch' grip. This grip can be used on a day-to-day basis because of its versatility, and provides the user with the functionality to live independently, with an improved quality of life.

The theoretical model of the orthosis was created using Matlab as a means to study the dynamics and kinematics of the orthosis. The solution was optimized using a built in Matlab function called 'fminsearch' and used an unconstrained non-linear search function based on the Nelder-Mead search algorithm. A program was created that was used with the fminsearch function, and when combined together, identified the attainable wrist position for a given set of orthosis geometries. The fminsearch function changed these geometries of the orthosis while trying to find the minimum attainable wrist position. The conclusion of the optimization process was that the fminsearch function found the local; rather than the global minimum of the orthosis. After testing different combinations it was found that the optimal design was approximately similar to the nearest positions to the surface of the skin as shown in Figure 11-1. This was desirable as it also simplified the concept and improved aesthetics.

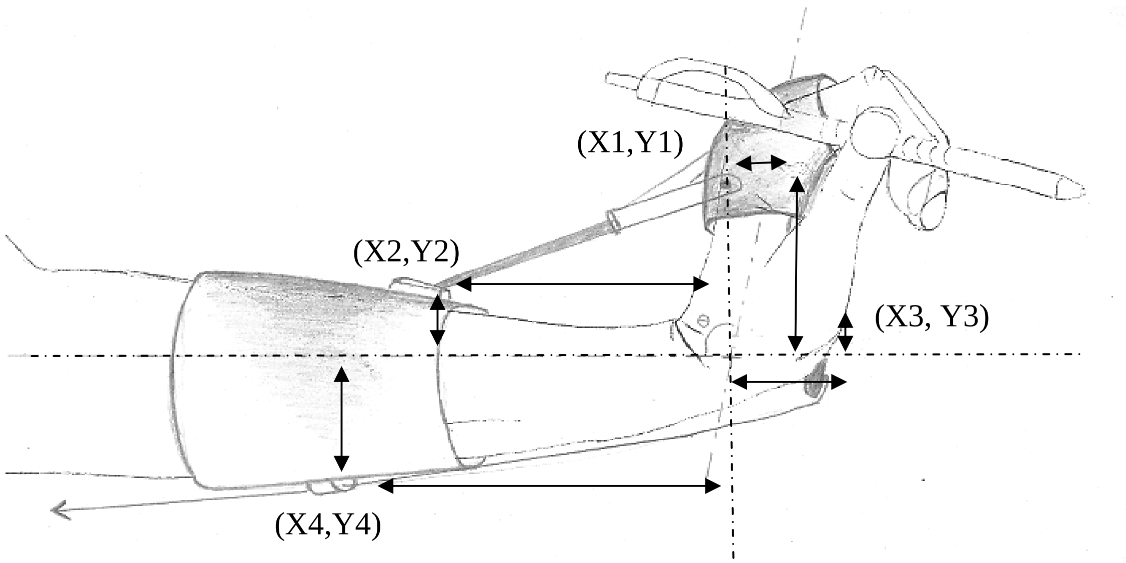


Figure 11-1 Example of the geometries of the orthosis being closely related to the positions nearest the surface of the skin

The orthosis was trialed with the help of two participants who were both high level tetraplegics, involved with rehabilitation at the Burwood Spinal Cord Unit. The aim was to compare the theoretical results of the orthosis to the experimental results when worn by the two participants. This analysis provided important information in regards to assumptions made during the construction of the theoretical model.

The functional outcome of the orthosis was predicted using a theoretical model which helped improve the design. The model simulated the required forces necessary to pull the wrist from 30° extension through to 30° flexion, and compared these results to the achievable forces from the shoulder. Using this analysis, the attainable angle of wrist rotation was determined at the point where the required force exceeded the achievable force. The limitations of this model included no account for friction, and data was collected from only two participants.

For both participants the theoretical model over-estimated the attainable wrist rotation when compared to the experimental results. This was partially caused by friction which not only acted on the cable and internally within the body joints, but also the support strap which tightened as the user elevated their shoulder. These two factors which were not accounted for in the theoretical model resulted in the low performance of the experimental results. Combined with this, the experimental trials demonstrated that the shoulder strengths used in the theoretical model were over-estimated. The shoulder strength used in the theoretical model was based on the maximum static achievable force. However, during the experimental trials it became apparent that the user could not generate this force repeatedly, hence making this measurement inappropriate for the model. Instead the dynamic forces of the shoulder should have been used, and rather than measuring the maximum dynamic forces, the user should work in a range which felt ‘comfortable’ on a day-to-day basis.

The orthosis succeeded in all but two design requirements which were the *range of motion* and the *speed of rotation.* These requirements were vital to the success of the orthosis and therefore further improvements were necessary to achieve them. An alternative wrist harness design was created which theoretically improved the functionality of the orthosis so that the requirements were achieved.

Chapter 12 Conclusion

12.1 Solution Overview

The designed solution was named the *self powered wrist extension orthosis*, more commonly referred to as the orthosis. The orthosis was separated into two sub-sections which included the shoulder harness and the wrist harness. The *shoulder harness* was remodeled from the design currently used with artificial upper body prosthetic limbs. It contains a brace which extends across the users back, which controls the wrist position as the user manipulates their shoulders. As the cable tightens the wrist is pulled into flexion and once released, a spring pulls the wrist back into extension. The mechanism surrounding the hand and wrist was called the *wrist harness* as shown in Figure 12-1.

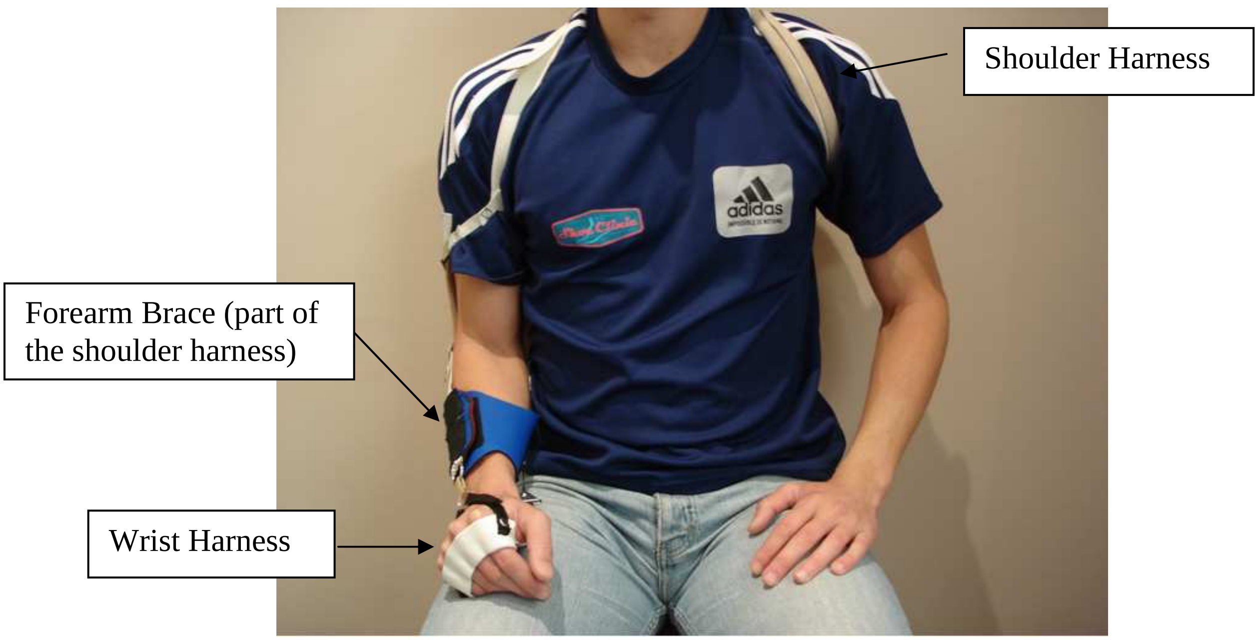


Figure 12-1 Anterior view of the orthosis

Figure 12-2 posterior view of the orthosis

12.2 Key Findings

1. The International Classification of Disability and Health (ICF) defined the design requirement specifications which detailed the need for the orthosis to be light, robust, and self-powered. Further requirements were a simple orthotic device, which was hidden under loose clothing and had the ability for the hand brace to be removed by the user when necessary, making the orthosis socially acceptable. The orthosis fulfilled the design requirement specifications identified in Chapter 3, apart from that it did not provide the user with 30° extension and 30° flexion which was critical for the success of this design.

2. Friction was not accounted for in the theoretical model; however it was later discovered that friction was present in the body joints, cabling system, and rubbing of the material. The approximated efficiency of the physical model was 70-80%, compared to the theoretical model.

3. The support strap which held the tricep pad in place was assumed to have no influence on the performance of the orthosis. However, during user trails it was found that as the user elevated their shoulder, the support strap consequently tightened, which in turn pulled on the tricep pad; increasing the spring extension. As a result, the user was required to

provide a greater force from the shoulder to overcome the spring stiffness which in turn reduced the attainable wrist rotation.

4. The shoulder strength used in the theoretical model was based on the maximum static achievable force, but during the experimental trials it became apparent that the user could not generate this force repeatedly, and hence this measurement was inappropriate for the model. Instead, the dynamic forces of the shoulder should have been measured, and rather than measuring the maximum dynamic forces, the user should work in a range which felt ‘comfortable’ on a day-to-day basis.

12.3 Recommendations

1. From concluding evidence, it is suggested that changing the geometries of the *current wrist harness* will be ineffective as it provides only a small functional increase but a significant decrease in aesthetics. Because the current orthosis design contains some inherent flaws it is recommended that this design be abandoned.

2. The theoretical results of the *alternative wrist harness* design provided significant improvement in both functionality and aesthetics. For this reason it is recommended that the alternative wrist harness be used for further research into a self powered wrist orthosis.

3. The figure-of-eight *shoulder harness*, which was used to power the *wrist harness*, provides energy which is both controlled and activated through a persons shoulder movement. This method is **highly recommended** because of the robustness, quitness, and ability to be concealed under clothing. Further development is necessary in maximizing the users output through the following:

    a. Reduce friction in the cables

    b. Reduce tension in the support strap through adding an elastic cross section

    c. Eliminate need of support strap through adapting a figure-of-nine shoulder harness design

d. Calculate shoulder strengths based on the users dynamic forces in a range which feels ‘comfortable to the user

4. Further development away from the wrist extension orthosis is recommended due to the large forces required to manipulate the wrist joint. It is therefore **strongly recommended** that the *shoulder harness* be combined with a *key pinch orthosis* to control the position of the thumb rather then controlling the position of the wrist. This combination requires less force from the shoulders and reduces the complexity of grasping objects.

5. Optional concepts such as the *motor drive* shown in Figure H.3 Appendix H, P165 should be considered. However any externally powered systems will have difficulty with control and activation, and it is therefore recommended that research into this system be conducted after eliminating all possibilities of self powered systems.

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Appendix A

Figure A.1 shows an example of generic grasp and release test which includes; a videotape, blocks, cylinder, pens, knife and fork.

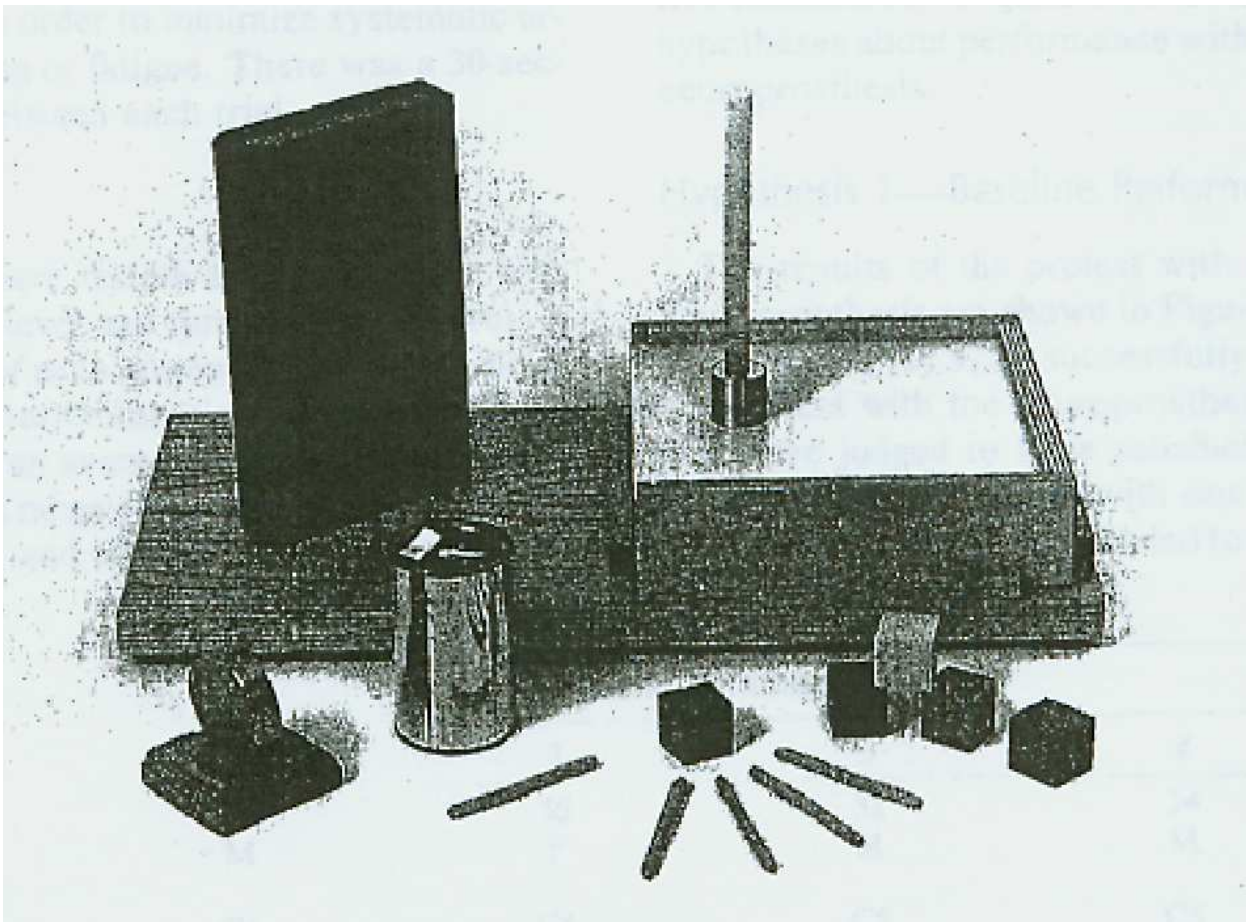


Figure A.1 The Grasp and Release Test (GRT) Apparatus (Wuolle and Doren, 1994)

Figure A.2 is the Functional Independence Measure (FIM) which is used to assess a person's execution of a task.

Self-Care	**Locomotion**	**Functional Independence Measure**
Eating	Walk/wheelchair	Level 7 Complete Independence
Grooming	stairs	
Bathing		Level 6 Modified Independence
Dressing - upper body	**Communication**	
Dressing - lower body	Comprehension	Level 5 Supervision Standby Assist
toileting	Expression	
		Level 4 Minimal Assist
Sphincter Control	**Social Cognition**	
Bladder Management	Social Interaction	Level 3 Moderate Assist
Bowel Management	problem Solving	
	Memory	Level 2 Maximal Assist
Transfers		
Bed, Chair, Wheelchair	**Total Score**	Level 1 Total Assist
Toilet	**Recorder**	
Tub, Shower		

Figure A.2 the Functional Independence Measure (FIM)

Figure A.3 shows an extract from the Impact on Participation and Autonomy (IPA) used to measure a person's social behavior.

Score: for office use only

Self care (with or without aids or assistance)

The next questions concern your personal care. When answering these questions, think about whether you can decide yourself when and how you want things done, even when you are assisted by someone else.

2a. My chances of getting washed and dressed *the way* I wish are

Very Good	☐	0
Good	☐	1
Fair	☐	2
Poor	☐	3
Very Poor	☐	4

2b. My chances of getting washed and dressed *when* I want to are

Very Good	☐	0
Good	☐	1
Fair	☐	2
Poor	☐	3
Very Poor	☐	4

2c. My chances of getting up and going to bed *when* I want to are

Very Good	☐	0
Good	☐	1
Fair	☐	2
Poor	☐	3
Very Poor	☐	4

2d. My chances of going to the toilet *when* I wish and need to are

Very Good	☐	0
Good	☐	1
Fair	☐	2
Poor	☐	3
Very Poor	☐	4

2e. My chances of eating and drinking *when* I want to are

Very Good	☐	0
Good	☐	1
Fair	☐	2
Poor	☐	3
Very Poor	☐	4

2f. If your health or your disability affect your self care, to what extent does this cause you problems?

No problems	☐	0
Minor problems	☐	1
Major problems	☐	2

Figure A.3 The Impact on Participation and Autonomy (IPA)

Appendix B

Arm brace designs

1. Figure B.1 was the most basic arm brace design, simply made from a single piece of aluminum plate with soft foam padding. This design was heavy and uncomfortable.

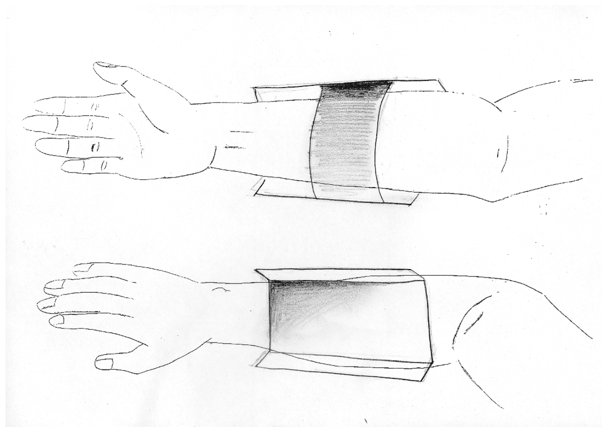


Figure B.1 Aluminium Arm Brace

2. Figure B.2 was a heat molded plastic arm brace which was chosen to avoid the need for the ‘tricep pad support straps’. This design was difficult to fit to the user.

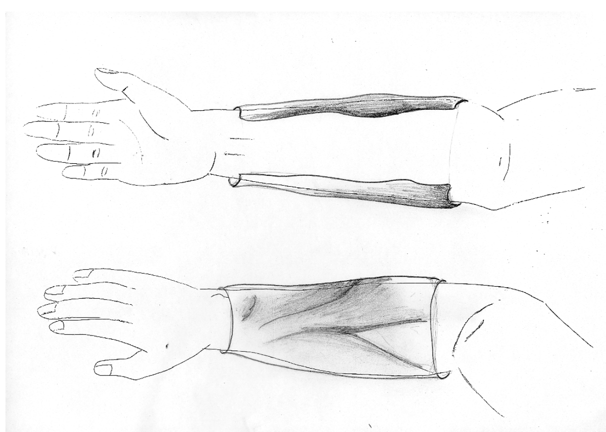


Figure B.2 Plastic Arm Brace

3. Figure B.3 shows a brace design which requires the ‘tricep pad support straps’ which eliminated the need for the brace to support x-component forces. The brace however was difficult to put on and twisted easily

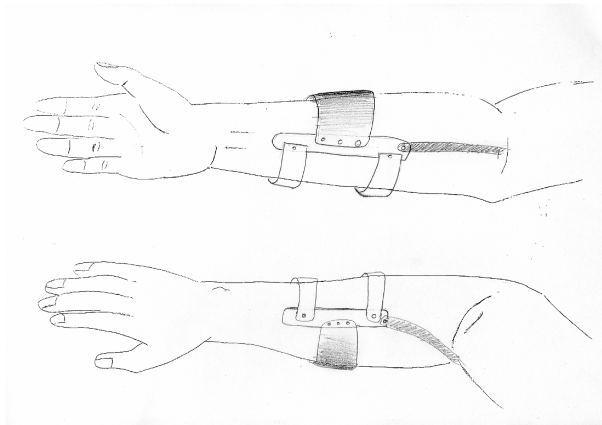


Figure B.3 Two Strap System

4. Figure B.4 was the final arm brace design which contained two metal inserts on the top and bottom and utilized the ‘tricep pad support straps’. It was simple to fit onto the user with the help of Velcro on both ends.

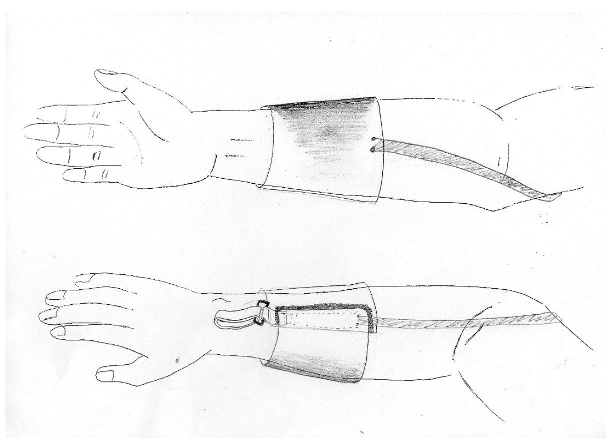


Figure B.4 Final Concept Design

Hand brace designs

1. Figure B.5 did not support x-component forces which resulted in the hand brace sliding.

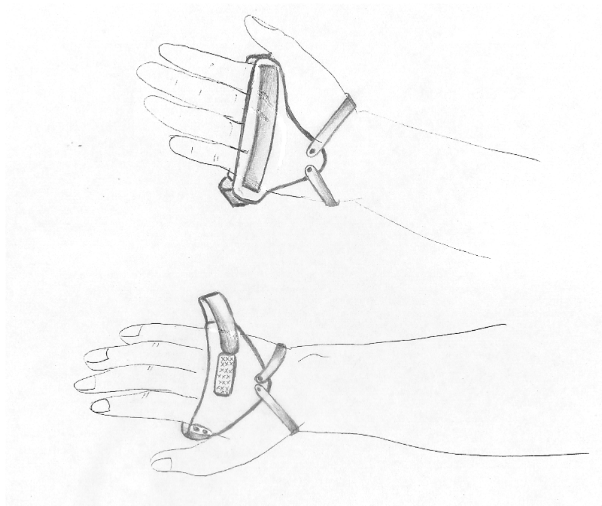


Figure B.5

2. Figure B.6 used leather straps in between the fingers which spread the x-component forces and stopped the hand brace from slipping. This design proved to be difficult to fit on the user, and created points of high pressure between the fingers.

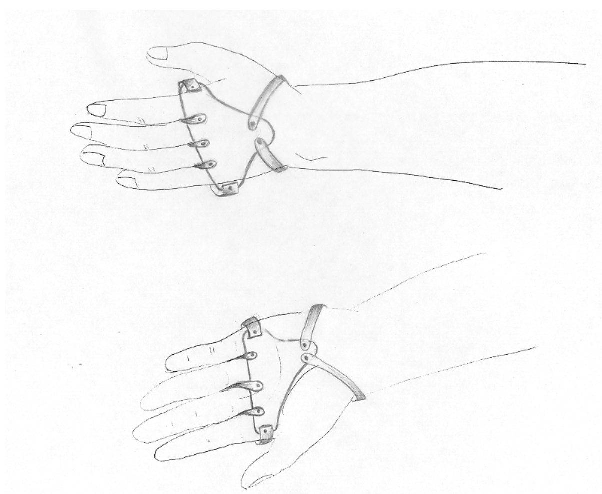


Figure B.6

3. The aim of Figure B.7 was to spread the load through the gate between the thumb and index finger. However the result was the leather sagged and caused a point of high pressure

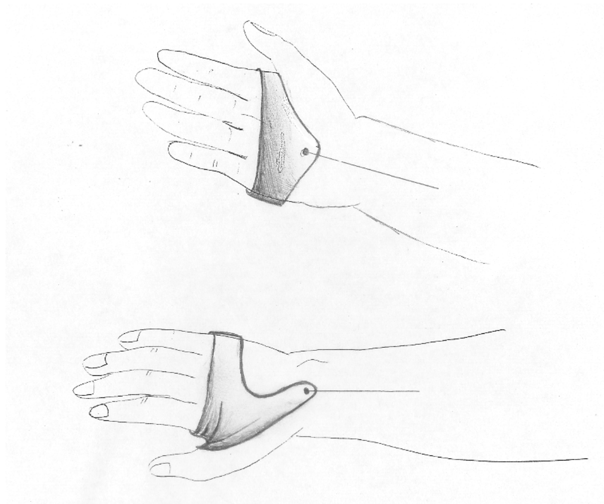


Figure B.7

4. The hand brace as shown in Figure B.8 was used in conjunction with the 'clock spring' design as shown in Appendix H.

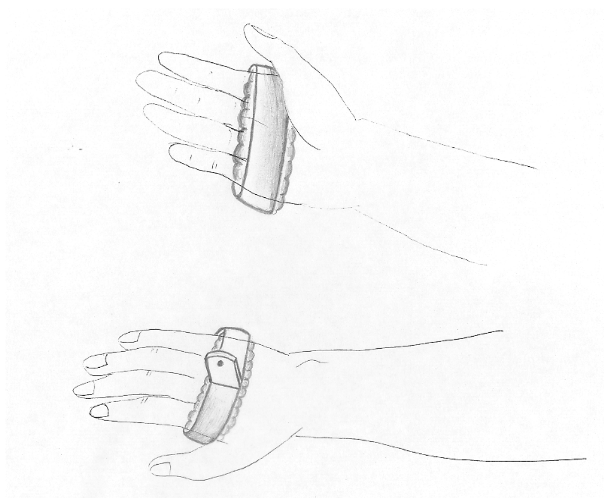


Figure B.8

5. Figure B.9 was a simple hand brace design, however it did not support x-component forces.

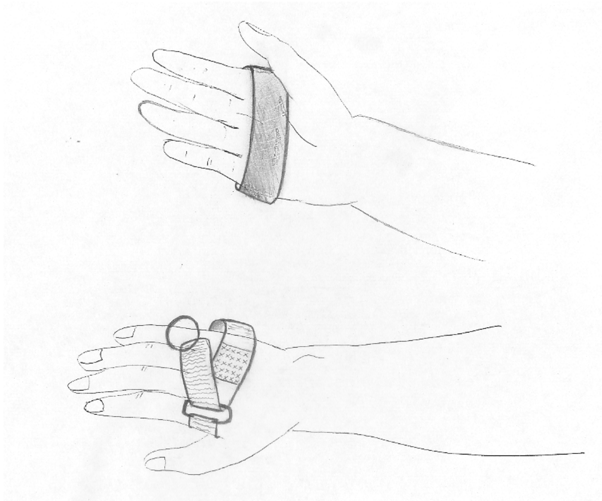


Figure B.9

6. Figure B.10 spread the load across the fingers, however the 'mitten' slipped over the fingers and did not achieve any wrist rotation.

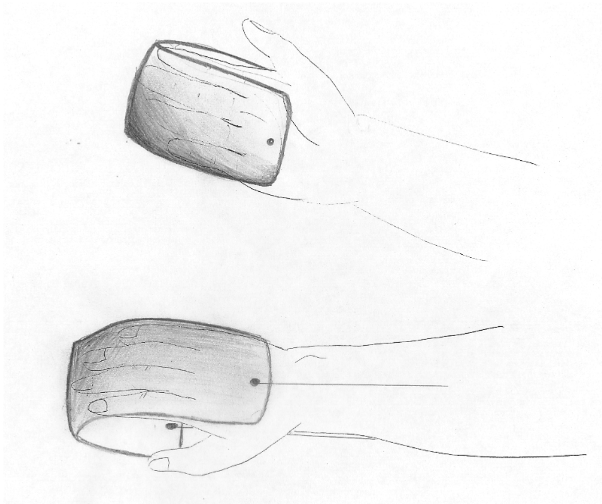


Figure B.10

7. Figure B.11 eventually became the successful concept. It spread the load over the palm of the hand while the wrist was in extension and spread the load across the fingers when the wrist was in flexion

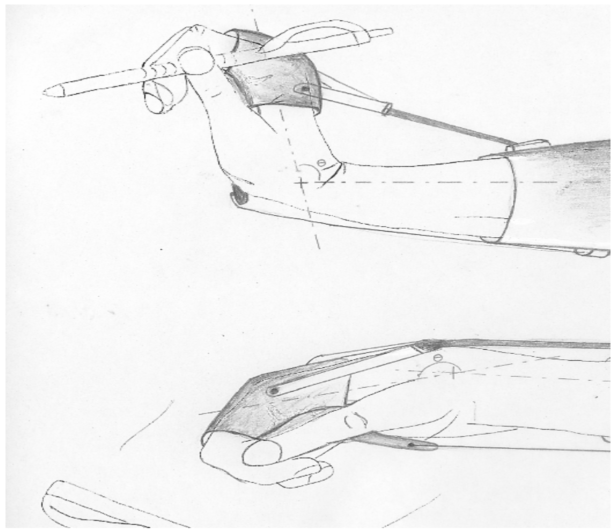


Figure B.11

Appendix C

Matlab code for the 'Current' orthosis Design.

```
%Shoulder orthosis Characteristics Created by Mathew Singer 30/11/05 All
%All dimensions are in meters
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%



%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%Upper Arm the stroke length takes into account slippage between the
%shoulder anchor and the back position. The slippage was assumed to only
%act on shoulder abduction and not flexion.
% clear all
% clc
% clf reset
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
function nothing = C1(X)
%X=[orthosis geometries]
%Maximum Extension
MaxE = 30;
%Maximum Flexion
MaxF = -30;
%WristPiv1 is the Heigth above the theoretical wrist pivot
WristPiv1 = .025;
%WristPiv2 is the Heigth below the theoretical wrist pivot
WristPiv2 = -.025;
%Shoulder Width
ShoulderWidth = 300/1000;
%Tricep position is when the arms are flat against the body, it was
%measured from the top of the shoulder down the upper arm
TricepPos = 80/1000;
Rotation = 90; %Degrees
Abduction = 85; %Degrees
Flexion =56; %Degrees
%Freedom is the angle in shoulder abduction and flexion where the arm is
%left unrestricted
FreedomAbd = 35; %Freedom in Degrees of Abuction
FreedomFle = 35; %Freedom in Degrees of Flexion
ForceAbd(1)=80; %Force in Abduction
ForceFle(1) =80; %Force in Flexion
%Input physical properties of the hand.
```

```
%Centre of Gravity from the wrist
HandCOG = .07;
%Mass of the hand
HandMass = .6; %Refer to workbook
%Object Mass is the expected max load
ObjectMass=0;
%Object Distance is to the key pinch from the wrist
ObjectCOG=0.05;
%Gravity
Gravity=9.81;
%Efficiency of the device from shoulder strap through to cable actuator.
Efficiency = 1;
%This option elinimates the need for the hand. 1 is true and 0 is false
Concept=1;
CentreLineX = [0];
CentreLineY = [0];


X1 = X(1);
X2 = X(2);
X3 = X(3);
X4 = X(4);
Y1 = X(5);
Y2 = X(6);
Y3 = X(7);
Y4 = X(8);
KElastic = 900

%INITIAL CALCULATION
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
RotCount=5; %Rotational devisions between Abduction and Flexion
DCount=Rotation/RotCount; %Number of iterations used
DiffF=FreedomFle-FreedomAbd; %Difference between Abduction angle and Flexion
PeakForce = 30; %Newton (N)
i=0;
Flexion2=0;
Abduction2=0;
AddA=5;%Add Abduction
Add=Abduction/AddA; %Add is the interval
AddF=Flexion/Add; %Add Flexion
RotationCapture=(Rotation/2)/RotCount %Captured data is at the mid point to compare to the required data
AbductionCapture=0  %Abduction Capture is needed to record the point where the achievable stroke length begins
DCountFle = ForceFle/Add; %Down Count from Flexion to zero for force calcs
DCountAbd = ForceAbd/Add; %Down Count from Abduction to zero for force calcs
OLength=sqrt(ShoulderWidth^2+TricepPos^2)
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
```

```
%CALCULATE SHOULDER ACHIEVABLE STROKE FORCE AND LENGTH
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%The following iterations are used to calculate the range of motion and
%range of forces achievable from the above specifications.
for Theta = 0:AddA:Abduction; %Achievable Shoulder Abduction
   j=0;
   i = i+1;
   Theta(i)=Theta;
   Diff = Flexion2(i)-Abduction2(i); %Force difference to calculate gradient between Flexion and Abduction
   ForceDiff(i)=ForceFle(i)-ForceAbd(i); %Force difference to calculate gradient between Force Flexion and Force Abduction
   %Rise, Run and m is used to calculate the back position as it changes
   %while the arm is raised from adduction to abduction.

   %The Rotation is Medial Rotation of the upper arm towards the flexed
   %position
   for Beta = 0:RotCount:Rotation; %Achievable Shoulder Flexion
      j=j+1;
      Beta2(i,j) = Beta;


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
      %The angle changes as the arm rotates from abduction to flexion.
      ThetaNew(i,j)=(Diff/Rotation)*Beta2(i,j)+Theta(i);

      %RX,RY,RZ are the local co-ordinates of the
      %arm!!!!!!!!!!!!!!!!!!!!!!!!
      RX(i,j)=TricepPos*sin(ThetaNew(i,j)*pi()/180);
      RY(i,j)=TricepPos*cos(ThetaNew(i,j)*pi()/180);
      RZ(i,j)=RX(i,j)*sin(Beta2(i,j)*pi()/180);

      %GX, GY, GZ are the global co-ordinates of the arm
      ThetaG(i,j)=(-Theta(1)/Rotation)*Beta2(i,j)+Theta(i);
      GX(i,j)=TricepPos*sin(ThetaG(i,j)*pi()/180);
      GY(i,j)=TricepPos-TricepPos*cos(ThetaNew(i,j)*pi()/180);
      GZ(i,j)=RX(i,j)*sin(Beta2(i,j)*pi()/180);

      TheoryStroke(i,j)=sqrt((RX(i,j)+ShoulderWidth)^2+(RY(i,j))^2)-OLength;


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
      %The following calculation determines the boundary of the "free"
      %area where the upper arm is free to move without being restrained
      %by the orthosis.
```

```
        %Free Relative Theta is the relative angle as the arm is rotated
        %from abduction through to flexion.
        ThetaRFree(i,j)=(DiffF/Rotation)*Beta2(i,j)+FreedomAbd;
        %Free Relative X Co-ordinate... this is needed to calculate the
        %global Z co-ordinate
        FRX(i,j)=TricepPos*sin(ThetaRFree(i,j)*pi()/180);
        FRY(i,j)=TricepPos*cos(ThetaRFree(i,j)*pi()/180);

        %The Global positions are calculated below
        ThetaGFree(i,j)=(-FreedomAbd/Rotation)*Beta2(i,j)+FreedomAbd;
        FGX(i,j)=TricepPos*sin(ThetaGFree(i,j)*pi()/180);
        FGY(i,j)=TricepPos-TricepPos*cos(ThetaRFree(i,j)*pi()/180);
        FGZ(i,j)=FRX(i,j)*sin(Beta2(i,j)*pi()/180);

        %The length of the free stroke
        FreeStroke(i,j)=sqrt((FRX(i,j)+ShoulderWidth)^2+(FRY(i,j))^2)-OLength;

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
        %The force changes as the arm rotates from abduction to flexion
        ForceNew(i,j)=(ForceDiff(i)/Rotation)*Beta2(i,j)+ForceAbd(i);



%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
        %The actual stroke is the difference between the theoretical stroke
        %and the area of free stroke.
        ActualStroke(i,j)=TheoryStroke(i,j)- FreeStroke(i,j);
        %If the actual stroke is less then zero then the co-ordinates of
        %the actual stroke are also set to zero
        if ActualStroke(i,j)<0;
            ActualStroke(i,j)=0;
            GX2(i,j)=0;
            GY2(i,j)=0;
            GZ2(i,j)=0;
            %If the actual stroke is greater is greater then zero then the
            %co-ordinates have the free co-ordinates taken off to give the
            %resulting true value

        else
            GX2(i,j)=GX(i,j)- FGX(i,j);
            GY2(i,j)=GY(i,j)- FGY(i,j);
            GZ2(i,j)=GZ(i,j)-FGZ(i,j);
            FreeForceNew(i,j)=ForceNew(i,j);
        end
        plot3(GX2(i,j),GZ2(i,j),GY2(i,j),':bs');
        hold on
    end
```

```
    %ForceFle is the Flexion Force. it is decremented each iterations such
    %that it equals zero on the last iteration. This is the same for
    %Abduction as well
    ForceFle(i+1) = ForceFle(i)-DCountFle;
    ForceAbd(i+1) = ForceAbd(i)-DCountAbd;

    %Abduction and flexion increases for each iteration until it reaches
    %the maximum value
    Abduction2(i+1)=Abduction2(i)+AddA;
    Flexion2(i+1)=Flexion2(i)+AddF;
end


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%The following calculations are necessary to find the mid point to compare
%values of the achievable shoulder outputs to the single value wrist
%outputs.
t=0
i3=0
for Theta = 0:AddA:Abduction;
    t=t+1;

    %Abduction Capture counts the number of times up to when the stroke
    %separates from zero. Check the stroke length graph to understand.
    if ActualStroke(t,RotationCapture)<=0;
        AbductionCapture=AbductionCapture+1;
    end

end

%The following section captures the average values of the stroke starting
%from the first non-zero value.
for i2 = AbductionCapture+1:i;
    i3=i3+1;
    AchievableStroke(i3) = ActualStroke(i2,RotationCapture);
    AchievableForce(i3)=FreeForceNew(i2,RotationCapture);
end
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%


%
% %PLOT
%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
% title(['Achievable stroke length with ' ,num2str(FreedomAbd),' degrees freedom abduction and ' ,num2str(FreedomFle),' degrees freedom flexion'])
% xlabel('X Axis')
% ylabel('Z Axis')
```

```
% zlabel('Y Axis')
% grid on
% axis tight
% hold off
%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
% % subplot(2,1,1)
% figure
% for i2 = 1:i;
%     for j2 = 1:j;
%         plot3(GX(i2,j2),GZ(i2,j2),GY(i2,j2),':bs');
%         hold on
%     end
% end
% grid on
% title({['Tricep Pad Co-ordinate Map'];
%     ['This model is based on a user who can achieve '  ,num2str(Abduction), ' degrees abduction and ' ,num2str(Flexion),' degrees flexion']})
% xlabel('x axis (m)')
% ylabel('y axis (m)')
% zlabel('z axis (m)')
% axis tight
% hold off
%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
% figure
% for i2 = 1:i;
%     for j2 = 1:j;
%         plot3(GX(i2,j2),GZ(i2,j2),ForceNew(i2,j2),':bs');
%         hold on
%     end
% end
% grid on
% title({['Predicted Achievable Force at Tricep Pad'];
%         ['based on user measurements of '   ,num2str(ForceAbd(1)), 'N abduction and ' ,num2str(ForceFle(1)),'N flexion while arm was in resting position']})
% xlabel('Abduction')
% ylabel('Flexion')
% zlabel('Force (N)')
% axis tight
% az = 55;
% el = 45;
% view(az, el);
% hold off
%
% %Plot of theoretical force using parameters from the Goniometer.
% figure
% % subplot(2,1,1)
```

```
% for i2 = 1:i;
%     for j2 = 1:j;
%         plot3(GX(i2,j2),GZ(i2,j2),FreeForceNew(i2,j2),':bs');
%         hold on
%     end
% end
% grid on
% title('Achievable Cable Force from shoulder over the free range of motion')
% xlabel('X Axis')
% ylabel('Y Axis')
% zlabel('Force (N)')
% axis tight
% az = 55;
% el = 45;
% view(az, el);
% hold off
%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%



%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
% %Plot of theoretical force using parameters from the Goniometer.
% % subplot(2,1,2)
% figure
% for i2 = 1:i;
%     for j2 = 1:j;
%         plot3(ThetaNew(i2,j2),Beta2(i2,j2),ActualStroke(i2,j2),':bs');
%         hold on
%     end
% end
% grid on
% title('Achievable Stroke length')
% xlabel('Shoulder Abduction (Degrees)')
% ylabel('Shoulder Rotation (Degrees)')
% zlabel('Stroke length (m)')
% axis tight
% hold off
%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
```

```
%ORTHOSIS CHARATERISTICS
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%(Optimal Design)
% if Concept == 1
%     Concept='Optimal Design'
%     %X1 Wrist to hand pivot x direction for spring
%     X1 = .02%.05;
%     %X2 Wrist to arm pivot x direction for spring
%     X2 = -.1028%-.1;
%     %X3 Wrist to hand pivot x direction for Cable
%     X3 = .12%.06;
%     %X4 Wrist to arm pivot x direction for Cable
%     X4 = -.0376%-0.04;
%     %Y1 Wrist to hand pivot y direction for spring
%     Y1 = .0227%.02;
%     %Y2 Wrist to arm pivot y direction for spring
%     Y2 = .0329%.03;
%     %Y3 Wrist to hand pivot y direction for cable
%     Y3 = -.0105%-.05;
%     %Y4 Wrist to arm pivot y direction for cable
%     Y4 = -.0661%-.1;
%     KElastic = 900;
%
%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%     %(Optimal Practical Design)
% elseif Concept == 2
%     Concept='Optimal Practical Design'
%     %X1 Wrist to hand pivot x direction for spring
%     X1 = .05;
%     %X2 Wrist to arm pivot x direction for spring
%     X2 = -.1;
%     %X3 Wrist to hand pivot x direction for Cable
%     X3 = .06;
%     %X4 Wrist to arm pivot x direction for Cable
%     X4 = -0.04;
%     %Y1 Wrist to hand pivot y direction for spring
%     Y1 = .02;
%     %Y2 Wrist to arm pivot y direction for spring
%     Y2 = .03;
%     %Y3 Wrist to hand pivot y direction for cable
%     Y3 = -.05;
%     %Y4 Wrist to arm pivot y direction for cable
%     Y4 = -.04;
%     KElastic = 900;
%
%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
```

```
%     %Current Design
% elseif Concept == 3
%     Concept='Current Concept Design'
%     %X1 Wrist to hand pivot x direction for spring
%     X1 = .07;
%     %X2 Wrist to arm pivot x direction for spring
%     X2 = -.07;
%     %X3 Wrist to hand pivot x direction for Cable
%     X3 = .055;
%     %X4 Wrist to arm pivot x direction for Cable
%     X4 = -0.13;
%     %Y1 Wrist to hand pivot y direction for spring
%     Y1 = .02;
%     %Y2 Wrist to arm pivot y direction for spring
%     Y2 = .035;
%     %Y3 Wrist to hand pivot y direction for cable
%     Y3 = -.02;
%     %Y4 Wrist to arm pivot y direction for cable
%     Y4 = -.04;
%     KElastic = 900;
%
% elseif Concept == 4
%     Concept='Fminsearch Optimised Design'
%     X1 =  0.0744
%     X2 = -0.0757
%     X3 = 0.0658
%     X4 = -0.1382
%     Y1 =0.0206
%     Y2 = 0.0333
%     Y3 = -0.0213
%     Y4 = -0.0442
%     KElastic = 223;
% end











%CALCULATION
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%Calculation for the spring length and Cable Length
i=0;
pop=0;
```

```
pop2=0;
figure

%Maximum Extension through to Maximum Flexion
for Theta = MaxE:-1:MaxF;
   i=i+1;
   Theta1(i) = Theta;

   Xnew2(i) = X2;
   Ynew2(i) = Y2;
   Xnew4(i) = X4;
   Ynew4(i) = Y4;


   %Calculates the position of the pivot respective to the hand and wrist
   Xnew1(i) = (X1*cos(Theta*pi()/180))-(Y1*sin(Theta*pi()/180));
   Ynew1(i) = (X1*sin(Theta*pi()/180))+(Y1*cos(Theta*pi()/180));
   Xnew3(i) = (X3*cos(Theta*pi()/180))-(Y3*sin(Theta*pi()/180));
   Ynew3(i) = (X3*sin(Theta*pi()/180))+(Y3*cos(Theta*pi()/180));

   %The Spring and Cable distances are calculated
   SpringY(i) = Ynew2(i)-Ynew1(i); %Difference in the Y
   SpringX(i) = Xnew2(i)-Xnew1(i); %Difference in the X
   CableY(i) = Ynew3(i)-Ynew4(i);
   CableX(i) = Xnew3(i)-Xnew4(i);

   %Check for collision at the wrist, if collision occurs c1 will be
   %lower then the upper wrist pivot and c2 will be above the lower wrist
   %pivot.
   m1(i) = SpringY(i)/SpringX(i);
   m2(i) = CableY(i)/CableX(i);
   c1(i) = Ynew1(i)-m1(i)*Xnew1(i);    %
   c2(i) = Ynew3(i)-m2(i)*Xnew3(i);    %

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%




%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
   %The next sections calculates if the spring collides with the wrist
   %therefor altering the elastic motion.
   %These calculations assume that the wrist moves as a pin joint which is
   %untrue=! However the calculations are more simple this way and this is
   %why it is done
   %There are seperate calculations depending on whether the spring is
   %contacting the wrist surface. If C1 is greater then the skin surface
```

```
%then there is no contact.
if c1(i) > WristPiv1;
   %Spring Length is the hypotanuse of the Y and X components of the
   %spring.
   SpringL(i) = sqrt(SpringY(i)^2+SpringX(i)^2);

   %Calculate the angle Omega for use to find the initial required
   %force to hold the hand in extension due to self weight. Omega is
   %the angle formed between the wrist and spring.
   %There are three sets of calculations necessary.
   if Ynew1(i)>Ynew2(i)
      Gamma(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
      Delta(i) = atan(abs(SpringY(i))/abs(SpringX(i)));
      Omega(i) = pi()/2 - Gamma(i) - Delta(i);
   elseif Ynew1(i)>0
      Gamma(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
      Delta(i) = atan(abs(SpringX(i))/abs(SpringY(i)));
      Omega(i) = pi() - Gamma(i) - Delta(i);
   else
      Gamma(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
      Delta(i) = atan(abs(SpringY(i))/abs(SpringX(i)));
      Omega(i) =  Gamma(i) + Delta(i) - pi()/2;
   end
   pop=0;



else
   %This calculation is used when the wrist interferes with the spring.
   %This calculation requires the spring to be broken up into two
   %sections.
   pop = pop + 1;

   %The spring is broken into two sections. The first section is the
   %length from the spring brace to the wrist pivot and the second
   %section is from the wrist pivot to the hand.
   SpringL1(i) = sqrt(Xnew2(i)^2+(Ynew2(i)-WristPiv1)^2);
   SpringL2(i) = sqrt(Xnew1(i)^2+(WristPiv1-Ynew1(i))^2);
   SpringL(i) = SpringL1(i)+SpringL2(i);

   %Calculate the angle Omega for use to find the initial required
   %force to hold the hand in extension due to self weight. Omega is
   %the angle formed between the wrist and spring.
   %There are three sets of calculations necessary.
   if Ynew1(i)>WristPiv1
      Gamma(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
      Delta(i) = atan(abs(Ynew1(i)-WristPiv1)/abs(Xnew1(i)));
      Omega(i) = pi()/2 - Gamma(i) - Delta(i);
```

```
        elseif Ynew1(i)>0
            Gamma(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
            Delta(i) = atan(abs(Xnew1(i))/(abs(Ynew1(i)-WristPiv1)));
            Omega(i) = pi() - Gamma(i) - Delta(i);
        else
            Gamma(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
            Delta(i) = atan(abs(Ynew1(i)-WristPiv1)/abs(Xnew1(i)));
            Omega(i) =  Gamma(i) + Delta(i) - pi()/2;
        end
    end

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%



%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
    %The following section calculates the force in the Spring assuming
    %linear behaviour of the spring. The initial force in the spring is
    %calculated from the weight of the hand and any object which the hand
    %is lifting. The spring is assumed to have zero extension at this stage.

    %Generated moment from the weight of the hand.
    HandMoment(i)=HandMass*Gravity*HandCOG*abs(cos(Theta*pi()/180));
    %Generated moment from the weight of an object
    ObjectMoment(i)=ObjectMass*Gravity*ObjectCOG*abs(cos(Theta*pi()/180));
    %The initial force is calculated by the minimum force required to lift
    %the weight of the hand and an object
    InitialForce(i)                                                   =
(ObjectMoment(i)+HandMoment(i))/((sqrt(Ynew1(i)^2+Xnew1(i)^2))*sin(Omega(i)));
    %Extension
    Extend(i) = SpringL(i)-SpringL(1);
    %The force in the elastic is a combination between the extension and
    %initial force.
    Force(i) = KElastic*Extend(i)+InitialForce(1); %Returns it to m from mm.


    %The following section brakes the elastic into its comoponents so the
    %moments can be calculated
    if pop==0;
        SpringForceX(i)=(SpringX(i)/SpringL(i))*Force(i);
        SpringMomentX(i) = -SpringForceX(i)*Ynew1(i);
        SpringForceY(i)=(SpringY(i)/SpringL(i))*Force(i);
        SpringMomentY(i) = SpringForceY(i)*Xnew1(i);
    else
        SpringForceX(i)=(Xnew1(i)/SpringL2(i))*Force(i);
        SpringMomentX(i) = SpringForceX(i)*Ynew1(i);
        SpringForceY(i)=((WristPiv1-Ynew1(i))/SpringL2(i))*Force(i);
        SpringMomentY(i) = SpringForceY(i)*Xnew1(i);
```

```
    end

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%



%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
    CableY2(i) = Ynew3(i)-WristPiv2;
    CableX2(i) = Xnew3(i);

    %The following section checks if the cable collides with the wrist and
    %alters the calculations..... refer to Appendices to see calculation
    %details.
    if c2(i) <= WristPiv2;
        CableL(i) = sqrt(CableY(i)^2+CableX(i)^2);

        if Ynew3(i) >= 0
            Psi(i) = atan(CableX(i)/CableY(i));
            Lamda(i) = atan(Xnew3(i)/Ynew3(i));
            Beta(i) = Lamda(i)-Psi(i);

        elseif Ynew3(i) > Ynew4(i)
            Psi(i) = atan(CableX(i)/CableY(i));
            Lamda(i) = atan(Xnew3(i)/abs(Ynew3(i)));
            Beta(i) = pi()-Lamda(i)-Psi(i);

        else
            Psi(i) = atan(CableX(i)/abs(CableY(i)));
            Lamda(i) = atan(Xnew3(i)/abs(Ynew3(i)));
            Beta(i) = Psi(i)-Lamda(i);

        end



    else
        pop2 = pop2 + 1;
        CableL1(i) = sqrt(Xnew4(i)^2+(Ynew4(i)-WristPiv2)^2);
        CableL2(i) = sqrt(Xnew3(i)^2+(WristPiv2-Ynew3(i))^2);
        CableL(i) = CableL1(i)+CableL2(i);
        if pop2==1;
            CableL3(i)= CableL(i);
            Theta4=Theta1(i);
        end

        if Ynew3(i)>=0
            Psi(i) = atan(CableX2(i)/CableY2(i));
```

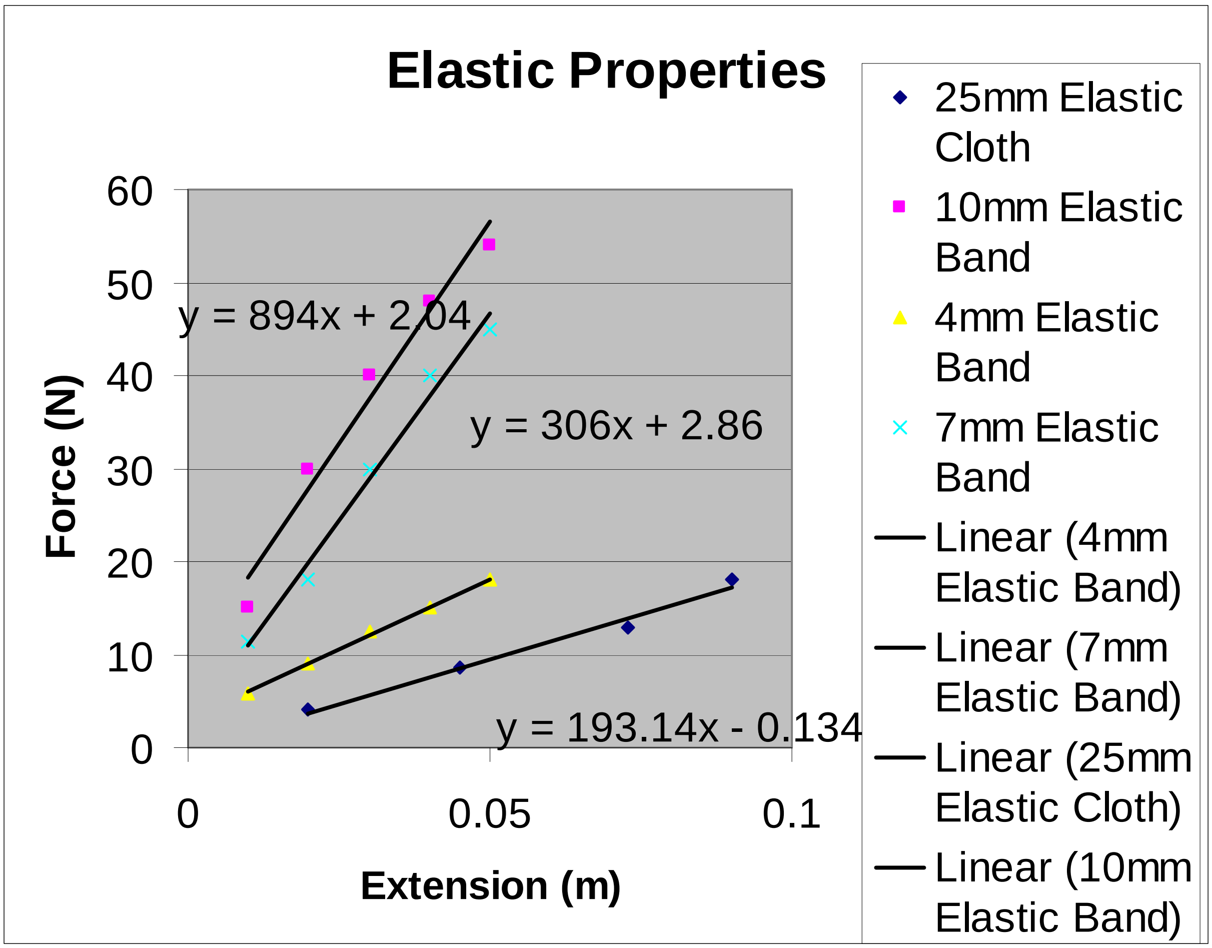
```
            Lamda(i) = atan(Xnew3(i)/Ynew3(i));
            Beta(i) = Lamda(i)-Psi(i);

        elseif Ynew3(i) > WristPiv2
            Psi(i) = atan(CableX2(i)/CableY2(i));
            Lamda(i) = atan(Xnew3(i)/abs(Ynew3(i)));
            Beta(i) = pi()-Lamda(i)-Psi(i);

        else
            Psi(i) = atan(abs(CableY2(i))/CableX2(i));
            Lamda(i) = atan(Xnew3(i)/abs(Ynew3(i)));
            Beta(i) = (pi()/2)-Lamda(i)-Psi(i);
        end


    end

    RequiredStroke(i) = (1/Efficiency)*(CableL(1)-CableL(i));
    RequiredForce(i) = (1/Efficiency)*(SpringMomentX(i)+SpringMomentY(i)-HandMoment(i)-
ObjectMoment(i))/(X3*sin(Beta(i)));

end



CaptureDiff=MaxE-MaxF;
CaptureDown=CaptureDiff/(i3-1);
i4=0;
for Theta = MaxE:-CaptureDown:MaxF;
    i4=i4+1;
    Theta4(i4) = Theta;
end


%PLOT
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%Achievable Force versus Required Force
%subplot(2,1,1)
figure
hold on
plot(Theta1,RequiredForce)
plot(Theta4,AchievableForce,':r')
set(gca,'XDir','reverse')
title('Required (solid) and Achievable (hashed) Cable Force to hold wrist statically');
%title('Required Cable Force to hold wrist statically at different positions of wrist extension');
%title('Achievable Cable Force');
ylabel('Cable Force (N)')
xlabel('Wrist Rotational Angle')
```

```
grid on
axis square
axis tight
hold off

%Achievable Stroke versus Required Stroke
%subplot(2,1,2)
figure
hold on
plot(Theta1,RequiredStroke)
plot(Theta4,AchievableStroke,':r')
set(gca,'XDir','reverse')
title('Required (solid) and Achievable (hashed) stroke lengths')
ylabel('Stroke Length (m)')
xlabel('Wrist Rotational Angle')
grid on
axis square
axis tight
hold off

BodyGeometriesX1 = [.035 WristPiv1 0.085 WristPiv2 -0.035];
BodyGeometriesY1= [-.2 0 .147 0 -.2];
BodyGeometriesX3 = [.035 WristPiv1 -0.085 WristPiv2 -0.035];
BodyGeometriesY3= [-.2 0 .147 0 -.2];

Xnew1A=[Xnew1(1) Xnew2];
Ynew1A=[Ynew1(1) Ynew2];

Xnew1B=[Xnew1(i) 0 Xnew2];
Ynew1B=[Ynew1(i) WristPiv1 Ynew2];

Xnew3A=[Xnew3(1) 0 Xnew4];
Ynew3A=[Ynew3(1) WristPiv2 Ynew4];

Xnew3B=[Xnew3(i) Xnew4];
Ynew3B=[Ynew3(i)  Ynew4];

%Overall orthosis dimensions
figure
%subplot(2,1,1)
hold on
title('Matlab Arm Wrist and Hand model in extended position')
plot(Xnew1A,Ynew1A,'b-+')
plot(Xnew3A,Ynew3A,'b-+')
plot(CentreLineX, CentreLineY,'--mo')
plot(BodyGeometriesY1,BodyGeometriesX1,'r+:')
ylabel('Y Position (m)')
xlabel('X Position (m)')
```

```
AXIS([-.2 .16 -.06 .1])
axis equal
hold off


% subplot(2,1,2)
% title('Matlab Arm Wrist and Hand model in flexed position')
% hold on
% plot(Xnew1B,Ynew1B,'b-+')
% plot(Xnew3B,Ynew3B,'b-+')
% plot(CentreLineX, CentreLineY,'--mo')
% plot(BodyGeometriesY3,BodyGeometriesX3,'r+:')
% ylabel('Y Position (m)')
% xlabel('X Position (m)')
% AXIS([-.2 .16 -.1 .06])
% axis equal
% hold off



figure
subplot(3,1,1)
plot(Theta1,SpringMomentX)
set(gca,'XDir','reverse')
title('Moment generated by the X Component of the Spring')
ylabel('Moment (Nm)')
xlabel('Rotational Angle (Degrees)')
axis square

subplot(3,1,2)
plot(Theta1,HandMoment)
set(gca,'XDir','reverse')
title('Moment generated by the weight of the hand')
ylabel('moment (Nm)')
xlabel('Rotational Angle (Degrees)')
axis square

subplot(3,1,3)
plot(Theta1,SpringMomentY)
set(gca,'XDir','reverse')
title('Moment generated by the Y Component of the Spring')
ylabel('Moment(Nm)')
xlabel('Rotational Angle (Degrees)')
axis square
```

Appendix D

Different spring properties



Appendix E

Matlab code for the ‘fminsearch’ optimization

```
%Shoulder orthosis Characteristics Created by Mathew Singer 30/11/05 All
%All dimensions are in meters
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

function MaxAngle = C1Opt(passd)
clear fun



%ORTHOSIS CHARATERISTICS

X1 = passd(1);
X2 = passd(2);
X3 = passd(3);
X4 = passd(4);
Y1 = passd(5);
Y2 = passd(6);
Y3 = passd(7);
Y4 = passd(8);
KElastic = 900;%passd(9);


%Maximum Extension
MaxE = 30;
%Maximum Flexion
MaxF = -30;
%WristPiv1 is the Heigth above the theoretical wrist pivot
WristPiv1 = .025;
%WristPiv2 is the Heigth below the theoretical wrist pivot
WristPiv2 = -.025;
%Shoulder Width
ShoulderWidth = 300/1000;
%Tricep position is when the arms are flat against the body, it was
%measured from the top of the shoulder down the upper arm
TricepPos = 80/1000;
Rotation = 90; %Degrees
Abduction = 85; %Degrees
Flexion =56; %Degrees
%Freedom is the angle in shoulder abduction and flexion where the arm is
%left unrestricted
```

```
FreedomAbd = 35; %Freedom in Degrees of Abuction
FreedomFle = 35; %Freedom in Degrees of Flexion
ForceAbd(1)=30; %Force in Abduction
ForceFle(1) =30; %Force in Flexion
%Input physical properties of the hand.
%Centre of Gravity from the wrist
HandCOG = .07;
%Mass of the hand
HandMass = .6; %Refer to workbook
%Object Mass is the expected max load
ObjectMass=0;
%Object Distance is to the key pinch from the wrist
ObjectCOG=0.05;
%Gravity
Gravity=9.81;
%Efficiency of the device from shoulder strap through to cable actuator.
Efficiency = 1;
%This option elinimates the need for the hand. 1 is true and 0 is false
Concept=4;
CentreLineX = [-.150 0 .150];
CentreLineY = [0 0 0];
R=.1;
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%


%Constraints on variables to lie within certain range
if (X1<0.11 & X1>0.009 & X2>-0.21 & X2<-0.009 & X3>0.009 & X3<0.11 & X4<-0.009 & X4>-0.21 & Y1<0.21 & Y1>0.015 & Y2<0.21 & Y2>0.035 & Y3>-0.21 & Y3<-0.015 & Y4>-0.21 & Y4<-0.035)

   %INITIAL CALCULATION

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

   RotCount=5; %Rotational devisions between Abduction and Flexion
   DCount=Rotation/RotCount; %Number of iterations used
   DiffF=FreedomFle-FreedomAbd; %Difference between Abduction angle and Flexion
   PeakForce = 30; %Newton (N)
   i=0;
   Flexion2=0;
   Abduction2=0;
   AddA=5;%Add Abduction
   Add=Abduction/AddA; %Add is the interval
   AddF=Flexion/Add; %Add Flexion
   RotationCapture=(Rotation/2)/RotCount; %Captured data is at the mid point to compare to the required data
```

```
   AbductionCapture=0;  %Abduction Capture is needed to record the point where the achievable stroke length begins
   DCountFle = ForceFle(1)/Add; %Down Count from Flexion to zero for force calcs
   DCountAbd = ForceAbd(1)/Add; %Down Count from Abduction to zero for force calcs
   OLength=sqrt(ShoulderWidth^2+TricepPos^2);

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%


   %CALCULATE SHOULDER ACHIEVABLE STROKE FORCE AND LENGTH

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
   %The following iterations are used to calculate the range of motion and
   %range of forces achievable from the above specifications.
   for Theta = 0:AddA:Abduction; %Achievable Shoulder Abduction
      j=0;
      i = i+1;
      Theta(i)=Theta;
      Diff = Flexion2(i)-Abduction2(i); %Force difference to calculate gradient between Flexion and Abduction
      ForceDiff(i)=ForceFle(i)-ForceAbd(i); %Force difference to calculate gradient between Force Flexion and Force Abduction
      %Rise, Run and m is used to calculate the back position as it changes
      %while the arm is raised from adduction to abduction.

      %The Rotation is Medial Rotation of the upper arm towards the flexed
      %position
      for Beta = 0:RotCount:Rotation; %Achievable Shoulder Flexion
         j=j+1;
         Beta2(i,j) = Beta;


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
         %The angle changes as the arm rotates from abduction to flexion.
         ThetaNew(i,j)=(Diff/Rotation)*Beta2(i,j)+Theta(i);

         %RX,RY,RZ are the local co-ordinates of the arm
         RX(i,j)=TricepPos*sin(ThetaNew(i,j)*pi()/180);
         RY(i,j)=TricepPos*cos(ThetaNew(i,j)*pi()/180);
         RZ(i,j)=RX(i,j)*sin(Beta2(i,j)*pi()/180);
         TheoryStroke(i,j)=sqrt((RX(i,j)+ShoulderWidth)^2+(RY(i,j))^2)-OLength;



%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
         %The following calculation determines the boundary of the "free"
```

```
            %area where the upper arm is free to move without being restrained
            %by the orthosis.

            ThetaRFree(i,j)=(DiffF/Rotation)*Beta2(i,j)+FreedomAbd;
            %Free Relative X Co-ordinate... this is needed to calculate the
            %global Z co-ordinate
            FRX(i,j)=TricepPos*sin(ThetaRFree(i,j)*pi()/180);
            FRY(i,j)=TricepPos*cos(ThetaRFree(i,j)*pi()/180);
            FreeStroke(i,j)=sqrt((FRX(i,j)+ShoulderWidth)^2+(FRY(i,j))^2)-OLength;


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
            %The force changes as the arm rotates from abduction to flexion
            ForceNew(i,j)=(ForceDiff(i)/Rotation)*Beta2(i,j)+ForceAbd(i);



%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
            %The actual stroke is the difference between the theoretical stroke
            %and the area of free stroke.
            ActualStroke(i,j)=TheoryStroke(i,j)- FreeStroke(i,j);
            %If the actual stroke is less then zero then the co-ordinates of
            %the actual stroke are also set to zero
            if ActualStroke(i,j)<0;
                ActualStroke(i,j)=0;
                %If the actual stroke is greater is greater then zero then the
                %co-ordinates have the free co-ordinates taken off to give the
                %resulting true value
            else
                FreeForceNew(i,j)=ForceNew(i,j);
            end

        end

        %ForceFle is the Flexion Force. it is decremented each iterations such
        %that it equals zero on the last iteration. This is the same for
        %Abduction as well
        ForceFle(i+1) = ForceFle(i)-DCountFle;
        ForceAbd(i+1) = ForceAbd(i)-DCountAbd;

        %Abduction and flexion increases for each iteration until it reaches
        %the maximum value
        Abduction2(i+1)=Abduction2(i)+AddA;
        Flexion2(i+1)=Flexion2(i)+AddF;
    end
```

```
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
   %The following calculations are necessary to find the mid point to compare
   %values of the achievable shoulder outputs to the single value wrist
   %outputs.
   t=0;
   for Theta = 0:AddA:Abduction;
      t=t+1;

      %Abduction Capture counts the number of times up to when the stroke
      %separates from zero. Check the stroke length graph to understand.
      if ActualStroke(t,RotationCapture)<=0;
         AbductionCapture=AbductionCapture+1;
      end

   end
   i3=0;
   %The following section captures the average values of the stroke starting
   %from the first non-zero value.
   for i2 = AbductionCapture+1:i;
      i3=i3+1;
      AchievableStroke(i3) = ActualStroke(i2,RotationCapture);
      AchievableForce(i3)=FreeForceNew(i2,RotationCapture);
   end

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%



   %WRIST OTHOSIS

   %CALCULATION

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
   %Calculation for the spring length and Cable Length
   i=0;
   pop=0;
   pop1=0;
   pop2=0;
   pop3=0;
   %Maximum Extension through to Maximum Flexion
   for Theta = MaxE:-R:MaxF;
      i=i+1;
      Theta1(i) = Theta;

      Xnew2(i) = X2;
      Ynew2(i) = Y2;
```

```
    Xnew4(i) = X4;
    Ynew4(i) = Y4;


    %Calculates the position of the pivot respective to the hand and wrist
    Xnew1(i) = (X1*cos(Theta*pi()/180))-(Y1*sin(Theta*pi()/180));
    Ynew1(i) = (X1*sin(Theta*pi()/180))+(Y1*cos(Theta*pi()/180));
    Xnew3(i) = (X3*cos(Theta*pi()/180))-(Y3*sin(Theta*pi()/180));
    Ynew3(i) = (X3*sin(Theta*pi()/180))+(Y3*cos(Theta*pi()/180));

    %The Spring and Cable distances are calculated
    SpringY(i) = Ynew2(i)-Ynew1(i); %Difference in the Y
    SpringX(i) = Xnew2(i)-Xnew1(i); %Difference in the X
    CableY(i) = Ynew3(i)-Ynew4(i);
    CableX(i) = Xnew3(i)-Xnew4(i);

    %Check for collision at the wrist, if collision occurs c1 will be
    %lower then the upper wrist pivot and c2 will be above the lower wrist
    %pivot.
    m1(i) = SpringY(i)/SpringX(i);
    m2(i) = CableY(i)/CableX(i);
    c1(i) = Ynew1(i)-m1(i)*Xnew1(i);    %
    c2(i) = Ynew3(i)-m2(i)*Xnew3(i);    %

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%




%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
    %The next sections calculates if the spring collides with the wrist
    %therefor altering the elastic motion.
    %These calculations assume that the wrist moves as a pin joint which is
    %untrue=! However the calculations are more simple this way and this is
    %why it is done
    %There are seperate calculations depending on whether the spring is
    %contacting the wrist surface. If C1 is greater then the skin surface
    %then there is no contact.
    if c1(i) > WristPiv1;
      %Spring Length is the hypotanuse of the Y and X components of the
      %spring.
      SpringL(i) = sqrt(SpringY(i)^2+SpringX(i)^2);

      %Calculate the angle Omega for use to find the initial required
      %force to hold the hand in extension due to self weight. Omega is
      %the angle formed between the wrist and spring.
      %There are three sets of calculations necessary.
```

```
    if Ynew1(i)>Ynew2(i)
        Gamma(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
        Delta(i) = atan(abs(SpringY(i))/abs(SpringX(i)));
        Omega(i) = pi()/2 - Gamma(i) - Delta(i);
    elseif Ynew1(i)>0
        Gamma(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
        Delta(i) = atan(abs(SpringX(i))/abs(SpringY(i)));
        Omega(i) = pi() - Gamma(i) - Delta(i);
    else
        Gamma(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
        Delta(i) = atan(abs(SpringY(i))/abs(SpringX(i)));
        Omega(i) =  Gamma(i) + Delta(i) - pi()/2;
    end
    pop=0;



else
    %This calculation is used when the wrist interferes with the spring.
    %This calculation requires the spring to be broken up into two
    %sections.
    pop = pop + 1;

    %The spring is broken into two sections. The first section is the
    %length from the spring brace to the wrist pivot and the second
    %section is from the wrist pivot to the hand.
    SpringL1(i) = sqrt(Xnew2(i)^2+(Ynew2(i)-WristPiv1)^2);
    SpringL2(i) = sqrt(Xnew1(i)^2+(WristPiv1-Ynew1(i))^2);
    SpringL(i) = SpringL1(i)+SpringL2(i);

    %Calculate the angle Omega for use to find the initial required
    %force to hold the hand in extension due to self weight. Omega is
    %the angle formed between the wrist and spring.
    %There are three sets of calculations necessary.
    if Ynew1(i)>WristPiv1
        Gamma(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
        Delta(i) = atan(abs(Ynew1(i)-WristPiv1)/abs(Xnew1(i)));
        Omega(i) = pi()/2 - Gamma(i) - Delta(i);
    elseif Ynew1(i)>0
        Gamma(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
        Delta(i) = atan(abs(Xnew1(i))/(abs(Ynew1(i)-WristPiv1)));
        Omega(i) = pi() - Gamma(i) - Delta(i);
    else
        Gamma(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
        Delta(i) = atan(abs(Ynew1(i)-WristPiv1)/abs(Xnew1(i)));
        Omega(i) =  Gamma(i) + Delta(i) - pi()/2;
    end
end
```

```
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%



%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
    %The following section calculates the force in the Spring assuming
    %linear behaviour of the spring. The initial force in the spring is
    %calculated from the weight of the hand and any object which the hand
    %is lifting. The spring is assumed to have zero extension at this stage.

    %Generated moment from the weight of the hand.
    HandMoment(i)=HandMass*Gravity*HandCOG*abs(cos(Theta*pi()/180));
    %Generated moment from the weight of an object
    ObjectMoment(i)=ObjectMass*Gravity*ObjectCOG*abs(cos(Theta*pi()/180));
    %The initial force is calculated by the minimum force required to lift
    %the weight of the hand and an object
    InitialForce(i) =
(ObjectMoment(i)+HandMoment(i))/((sqrt(Ynew1(i)^2+Xnew1(i)^2))*sin(Omega(i)));
    %Extension
    Extend(i) = SpringL(i)-SpringL(1);
    %The force in the elastic is a combination between the extension and
    %initial force.
    Force(i) = KElastic*Extend(i)+InitialForce(1); %Returns it to m from mm.


    %The following section brakes the elastic into its comoponents so the
    %moments can be calculated
    if pop==0;
      SpringForceX(i)=(SpringX(i)/SpringL(i))*Force(i);
      SpringMomentX(i) = -SpringForceX(i)*Ynew1(i);
      SpringAchievableForce(i)=(SpringY(i)/SpringL(i))*Force(i);
      SpringMomentY(i) = SpringAchievableForce(i)*Xnew1(i);
    else
      SpringForceX(i)=(Xnew1(i)/SpringL2(i))*Force(i);
      SpringMomentX(i) = SpringForceX(i)*Ynew1(i);
      SpringAchievableForce(i)=((WristPiv1-Ynew1(i))/SpringL2(i))*Force(i);
      SpringMomentY(i) = SpringAchievableForce(i)*Xnew1(i);
    end

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%



%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
    CableY2(i) = Ynew3(i)-WristPiv2;
```

```
CableX2(i) = Xnew3(i);

%The following section checks if the cable collides with the wrist and
%alters the calculations..... refer to Appendices to see calculation
%details.
if c2(i) <= WristPiv2;
   CableL(i) = sqrt(CableY(i)^2+CableX(i)^2);

   if Ynew3(i) >= 0
      Psi(i) = atan(CableX(i)/CableY(i));
      Lamda(i) = atan(Xnew3(i)/Ynew3(i));
      Beta(i) = Lamda(i)-Psi(i);

   elseif Ynew3(i) > Ynew4(i)
      Psi(i) = atan(CableX(i)/CableY(i));
      Lamda(i) = atan(Xnew3(i)/abs(Ynew3(i)));
      Beta(i) = pi()-Lamda(i)-Psi(i);

   else
      Psi(i) = atan(CableX(i)/abs(CableY(i)));
      Lamda(i) = atan(Xnew3(i)/abs(Ynew3(i)));
      Beta(i) = Psi(i)-Lamda(i);

   end



else
   pop2 = pop2 + 1;
   CableL1(i) = sqrt(Xnew4(i)^2+(Ynew4(i)-WristPiv2)^2);
   CableL2(i) = sqrt(Xnew3(i)^2+(WristPiv2-Ynew3(i))^2);
   CableL(i) = CableL1(i)+CableL2(i);
   if pop2==1;
      CableL3(i)= CableL(i);
      Theta4=Theta1(i);
   end

   if Ynew3(i)>=0
      Psi(i) = atan(CableX2(i)/CableY2(i));
      Lamda(i) = atan(Xnew3(i)/Ynew3(i));
      Beta(i) = Lamda(i)-Psi(i);

   elseif Ynew3(i) > WristPiv2
      Psi(i) = atan(CableX2(i)/CableY2(i));
      Lamda(i) = atan(Xnew3(i)/abs(Ynew3(i)));
      Beta(i) = pi()-Lamda(i)-Psi(i);

   else
```

```
            Psi(i) = atan(abs(CableY2(i))/CableX2(i));
            Lamda(i) = atan(Xnew3(i)/abs(Ynew3(i)));
            Beta(i) = (pi()/2)-Lamda(i)-Psi(i);
        end


    end

    RequiredStroke(i) = (1/Efficiency)*(CableL(1)-CableL(i));

    RequiredForce(i)          =          (1/Efficiency)*(SpringMomentX(i)+SpringMomentY(i)-
HandMoment(i)-ObjectMoment(i))/(X3*sin(Beta(i)));

end

%The following algorithm firstly finds the angle when the required
%force exceeds the achievable force. Using this result the angle is
%then compared to the stroke length. If the required stroke length is
%greater then the achievable then the new angle is found by the
%intersection of the achieavable stroke length at that angle on the
%required stroke length graph.

%Theta4 is the angle over which the captured stroke and force is
%calculated
b=0;
CaptureDiff=MaxE-MaxF;
CaptureDown=CaptureDiff/(i3-1);
i4=0;
for Theta = MaxE:-CaptureDown:MaxF;
    i4=i4+1;
    Theta4(i4) = Theta;
end


Theta5=Theta1*-1;
n=0;
k=0;
%loop over achievable and required force until the required force
%exceeds the achievable.
for k=MaxF:R:MaxE;
    n=n+1;
    % The cable force must be greater then zero otherwise the system is
    % unstable and will collapse. If the cable force is less then zero
    % make MaxAngle=30.
    if RequiredForce(n) >= 0
        %find corresponding angle where required force exceeds
        %achievable.
        pop3=0;
```

```
        m = ((AchievableForce(i4)-AchievableForce(1))/(Theta4(1)-Theta4(i4)));
        c = AchievableForce(1)-m*Theta5(1);
        PtAchievableForce = m*Theta5(n)+c;
        %Checks when the Cable force exceeds the achievable force.
        if PtAchievableForce < RequiredForce(n);
            MaxAngle = -Theta5(n);
            m2 = ((AchievableStroke(i4)-AchievableStroke(1))/(Theta4(1)-Theta4(i4)));
            c2 = AchievableStroke(1)-m2*Theta5(1);
            PtAchievableStroke = m2*Theta5(n)+c2;
            %Checks when the Cable stroke exceeds the achievable stroke.
            if PtAchievableStroke < RequiredStroke(n);
                %if the achievable stroke is less then the required
                %stroke then the new angle is calculated.
                [R,C] = find(RequiredStroke <= PtAchievableStroke);
                C=max(C);
                MaxAngle = Theta1(C);
                break
            else
                break
            end
        end
        %if the required force never exceeds the achievable force then
        %the system checks the stroke length parameters.
        if n == (MaxE-MaxF)*1/R;
            MaxAngle = MaxF;
            m2 = ((AchievableStroke(i4)-AchievableStroke(1))/(Theta4(1)-Theta4(i4)));
            c2 = AchievableStroke(1)-m2*Theta5(1);
            PtAchievableStroke = m2*Theta5(n)+c2;
            if PtAchievableStroke < RequiredStroke(n);
                [R,C] = find(RequiredStroke <= PtAchievableStroke);
                C=max(C);
                MaxAngle = Theta1(C);
                break
            end

        end
    else

        pop3=1;
        break
    end
end

% bravo=MaxAngle;
if pop3 == 1
    MaxAngle=30;
end
```

```
    %From the beginning if the orthosis requirements
else
    MaxAngle=30;
end
```

Appendix F

Hand Brace

The hand brace (Figure) was constructed out of a single piece of plastic which was molded to fit the user's hand. Two Control Straps were necessary which attached to the spring. The control straps were used to adjust the medial and lateral movement of the hand. The Cable Connector attached the hand brace to the cable and also adjusted the pre-tension in the cable.

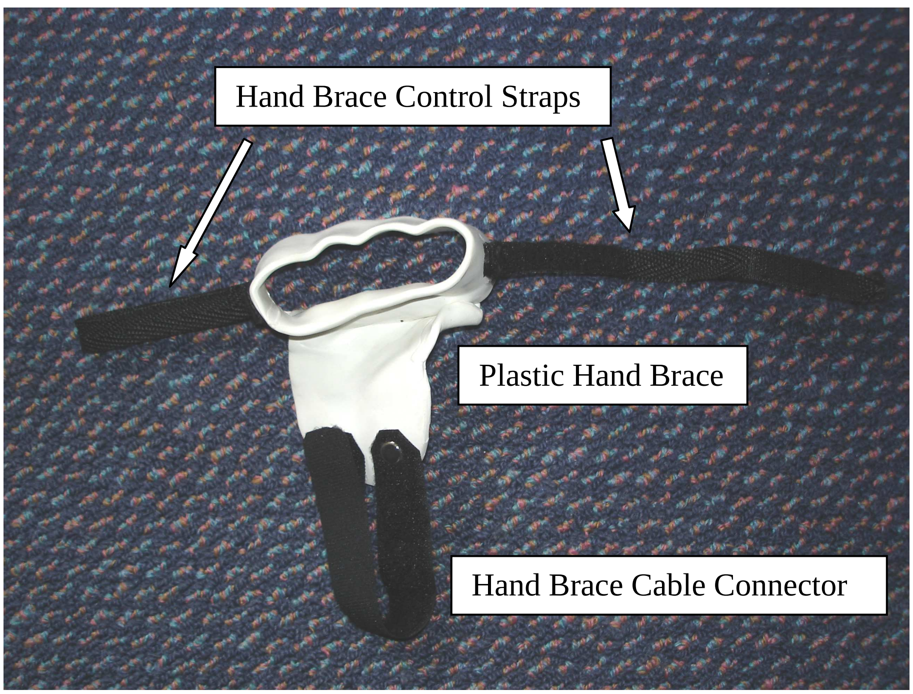


Figure F.1 Hand Brace Design

Arm Brace

The arm brace was made from a single piece of neoprene with Velcro patches on each end used to attach the two ends of the neoprene together. The Support Straps were made from 10mm Dacron fabric and were riveted to the aluminium inserts. The Spring was attached to the aluminium insert through a section of 25mm Dacron fabric. The Cable Anchor was used to attach the Distal Retainer on the Bowden Cable to the lower surface of the arm brace.

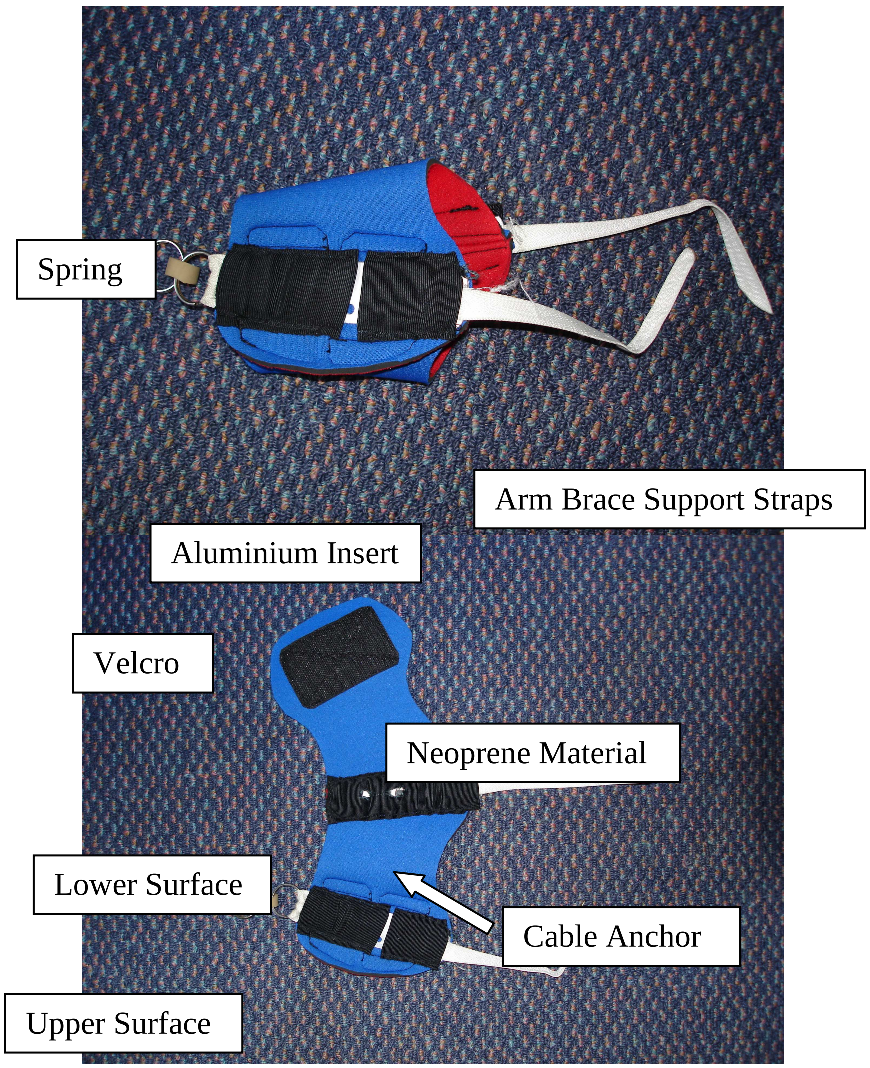


Figure F.2 Arm Brace

Tricep Pad

The Tricep Pad was constructed from leather 4mm thick. The Tricep Pad provided support to the Arm Brace through the Support Straps and provided an anchor for the Bowden Cable. The Tricep Pad was supported by the Y-Suspensor Support Straps which connected to the Shoulder Harness.

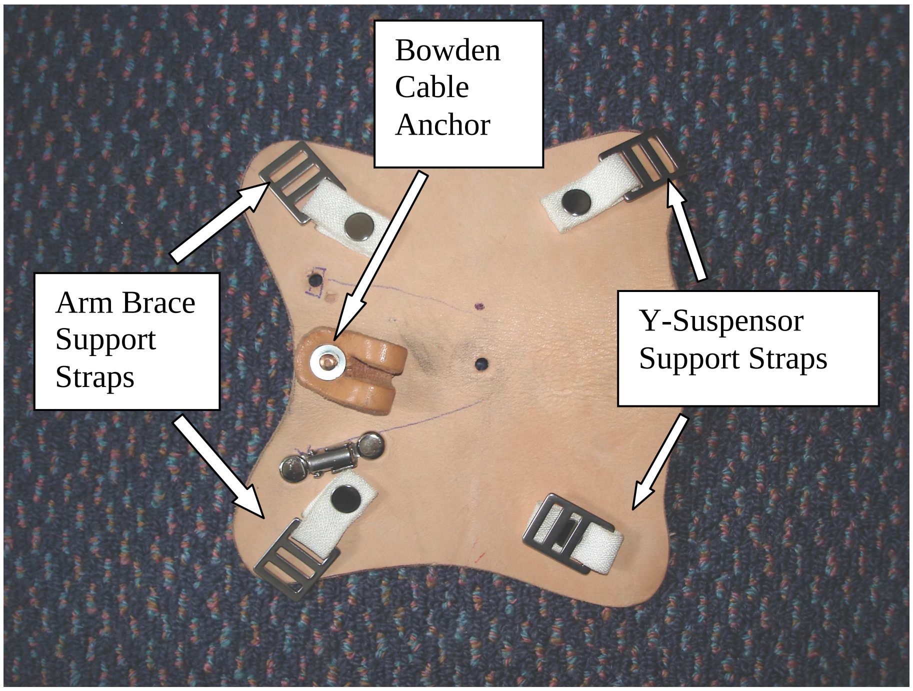


Figure F.3 Tricep Pad

Inverted Y Suspensor

The Inverted Y Suspensor connected to the shoulder and provided support to the Tricep Pad. The larger section of the Inverted Y Suspensor was made from 15mm Dacron and the smaller V-Section was made from a single piece of 10mm Dacron.

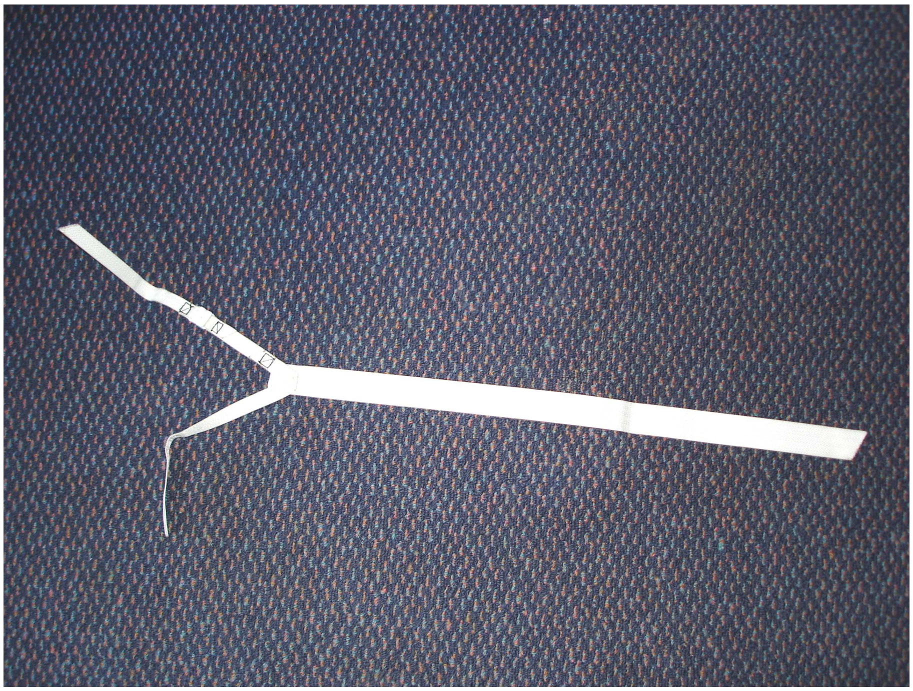


Figure F.4 Y Suspensor

Control Adjustment Strap

The Control Adjustment Strap which attached to the Bowden Cable transferred the energy from the shoulder and arm through to the wrist. The Distal Hanger connected to the Hand Brace while the Distal Retainer connected to the arm brace. The Proximal Hanger connected to the 25mm Dacron cloth and the Proximal Retainer connected to the Tricep Pad. The Distal and Proximal Retainers could both adjust along the Cable Sheath which was constructed from a cylindrical flexible metal. The internal inner cable was made from stainless steel. The cable and sheath was collectively called the Bowden Cable.

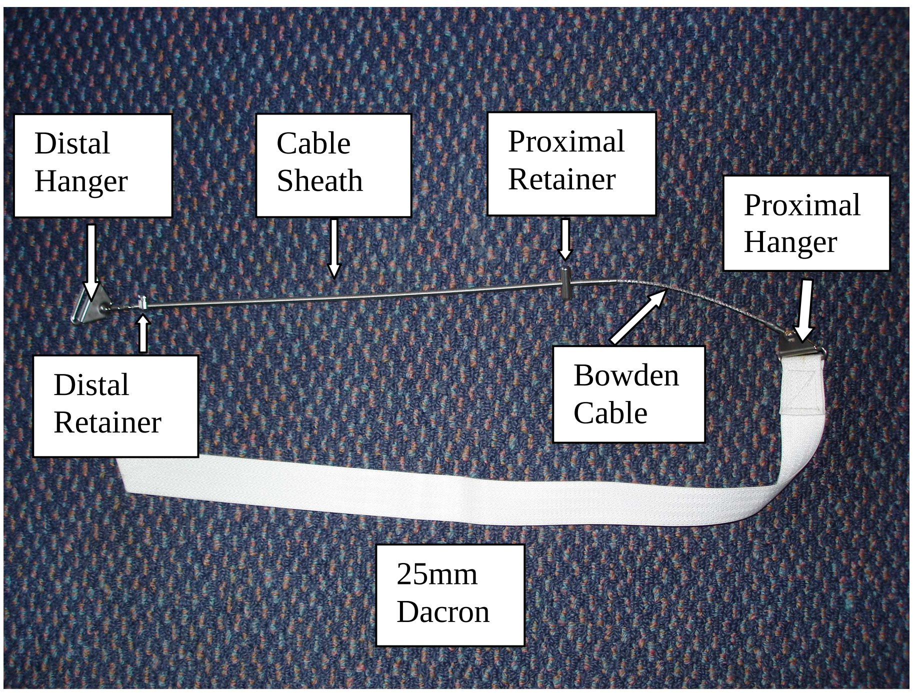


Figure F.5 Control Adjustment Strap

Auxiliary Shoulder Harness

The shoulder harness was necessary to support the Wrist orthosis. The padding was necessary to reduce the stress on the auxiliary shoulder and the buckle made the harness easier to put on and take off.

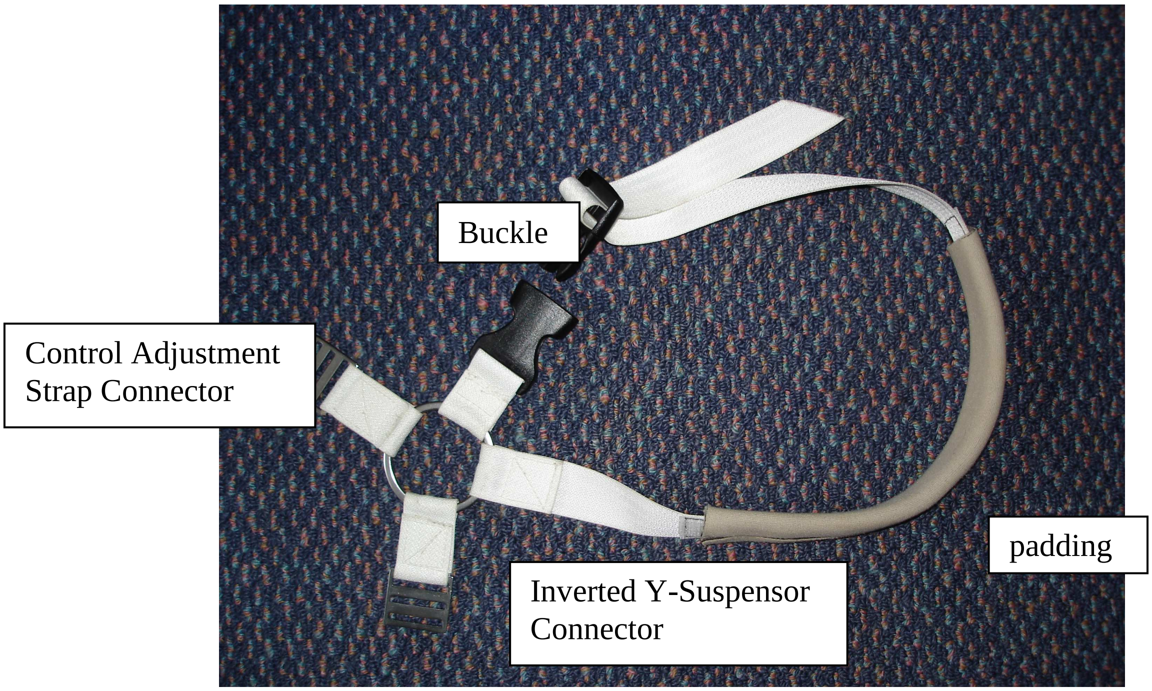


Figure F.6 Auxiliary Shoulder Harness

Appendix G

Matlab code for the alternative wrist harness design.

```
%Shoulder orthosis Characteristics Created by Mathew Singer 30/11/05 All
%All dimensions are in meters
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%Upper Arm
The stroke length takes into account slippage between the
%shoulder anchor and the back position. The slippage was assumed to only
%act on shoulder abduction and not flexion.
function C2(X)
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%


%PARAMETER INPUTS
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%The shoulder width was necessary to calculate the
%position where the cable detached from the back
ShoulderWidth = 300/1000;
%Tricep position is when the arms are flat against the body, it was
%measured from the top of the shoulder down the upper arm
TricepPos = 180/1000;
Rotation = 90; %Degrees
Abduction =85; %Degrees
Flexion =56; %Degrees
%Freedom is the angle in shoulder abduction and flexion where the arm is
%left unrestricted
FreedomAbd = 30; %Freedom in Degrees of Abuction
FreedomFle = 30; %Freedom in Degrees of Flexion
ForceAbd(1)=30; %Force in Abduction
ForceFle(1) =30; %Force in Flexion
FixedBackPos = 0 %1=True 0=False i.e. back anchor position changes with abduction
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%WRIST OTHOSIS
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%Input physical properties of the hand.
%Centre of Gravity from the wrist
```

```
HandCOG = .07;
%Mass of the hand
HandMass = .6; %Refer to workbook
%Object Mass is the expected max load
ObjectMass=0;
%Object Distance is to the key pinch from the wrist
ObjectCOG=0.05;
%Gravity
Gravity=9.81;
%Efficiency of the device from shoulder strap through to cable actuator.
Efficiency = 1;
%This option elinimates the need for the hand. 1 is true and 0 is false
HandEffect = 1;
%The Centreline is the theoretical pivotal positions.
CentreLineX = [0 0 0];
CentreLineY = [-.050 0 .050];
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
MaxE = 30;
%Maximum Flexion
MaxF = -30;
%WristPiv1 is the Heigth above the theoretical wrist pivot
WristPiv1 = .025;
%WristPiv2 is the Heigth below the theoretical wrist pivot
WristPiv2 = -.025;
KSpring=9000; %Spring constant N/m
%This result was taken from a 7mm rubber band


   %X1 Wrist to hand pivot x direction for spring
   X1 = .07;
   %X2 Wrist to arm pivot x direction for spring
   X2 = -.04;
   %X3 Wrist to hand pivot x direction for Cable
%    X3 = .055;
%    %X4 Wrist to arm pivot x direction for Cable
%    X4 = -0.13;
%    %Y1 Wrist to hand pivot y direction for spring
   Y1 = .02;
   %Y2 Wrist to arm pivot y direction for spring
   Y2 = .04;
%    %Y3 Wrist to hand pivot y direction for cable
%    Y3 = -.02;
%    %Y4 Wrist to arm pivot y direction for cable
%    Y4 = -.04;
```

```
%INITIAL CALCULATION
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
RotCount=5; %Rotational devisions between Abduction and Flexion
DCount=Rotation/RotCount; %Number of iterations used

DiffF=FreedomFle-FreedomAbd; %Difference between Abduction angle and Flexion
PeakForce = 30; %Newton (N)
i=0;
Flexion2=0;
Abduction2=0;
AddA=5;%Add Abduction
Add=Abduction/AddA; %Add is the interval
AddF=Flexion/Add; %Add Flexion
RotationCapture=(Rotation/2)/RotCount %Captured data is at the mid point to compare to the required data
AbductionCapture=0  %Abduction Capture is needed to record the point where the achievable stroke length begins
DCountFle = ForceFle/Add; %Down Count from Flexion to zero for force calcs
DCountAbd = ForceAbd/Add; %Down Count from Abduction to zero for force calcs
OLength=sqrt(ShoulderWidth^2+TricepPos^2)
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%



%CALCULATE SHOULDER ACHIEVABLE STROKE FORCE AND LENGTH
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%The following iterations are used to calculate the range of motion and
%range of forces achievable from the above specifications.
for Theta = 0:AddA:Abduction; %Achievable Shoulder Abduction
   j=0;
   i = i+1;
   Theta(i)=Theta;
   Diff = Flexion2(i)-Abduction2(i); %Force difference to calculate gradient between Flexion and Abduction
   ForceDiff(i)=ForceFle(i)-ForceAbd(i); %Force difference to calculate gradient between Force Flexion and Force Abduction
   %Rise, Run and m is used to calculate the back position as it changes
   %while the arm is raised from adduction to abduction.

   %The Rotation is Medial Rotation of the upper arm towards the flexed
   %position
   for Beta = 0:RotCount:Rotation; %Achievable Shoulder Flexion
      j=j+1;
      Beta2(i,j) = Beta;
```

```
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%The angle changes as the arm rotates from abduction to flexion.
ThetaNew(i,j)=(Diff/Rotation)*Beta2(i,j)+Theta(i);

%RX,RY,RZ are the local co-ordinates of the
%arm!!!!!!!!!!!!!!!!!!!!!!!!
RX(i,j)=TricepPos*sin(ThetaNew(i,j)*3.16/180);
RY(i,j)=TricepPos*cos(ThetaNew(i,j)*3.16/180);
RZ(i,j)=RX(i,j)*sin(Beta2(i,j)*3.16/180);

%GX, GY, GZ are the global co-ordinates of the arm
ThetaG(i,j)=(-Theta(1)/Rotation)*Beta2(i,j)+Theta(i);
GX(i,j)=TricepPos*sin(ThetaG(i,j)*3.16/180);
GY(i,j)=TricepPos-TricepPos*cos(ThetaNew(i,j)*3.16/180);
GZ(i,j)=RX(i,j)*sin(Beta2(i,j)*3.16/180);

TheoryStroke(i,j)=sqrt((RX(i,j)+ShoulderWidth)^2+(RY(i,j))^2)-OLength;

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%The following calculation determines the boundary of the "free"
%area where the upper arm is free to move without being restrained
%by the orthosis.

%Free Relative Theta is the relative angle as the arm is rotated
%from abduction through to flexion.
ThetaRFree(i,j)=(DiffF/Rotation)*Beta2(i,j)+FreedomAbd;
%Free Relative X Co-ordinate... this is needed to calculate the
%global Z co-ordinate
FRX(i,j)=TricepPos*sin(ThetaRFree(i,j)*3.16/180);
FRY(i,j)=TricepPos*cos(ThetaRFree(i,j)*3.16/180);

%The Global positions are calculated below
ThetaGFree(i,j)=(-FreedomAbd/Rotation)*Beta2(i,j)+FreedomAbd;
FGX(i,j)=TricepPos*sin(ThetaGFree(i,j)*3.16/180);
FGY(i,j)=TricepPos-TricepPos*cos(ThetaRFree(i,j)*3.16/180);
FGZ(i,j)=FRX(i,j)*sin(Beta2(i,j)*3.16/180);

%The length of the free stroke
%FreeStroke(i,j)=sqrt(FGZ(i,j)^2+(sqrt((FGX(i,j))^2+(FGY(i,j)-TricepPos3)^2))^2);
%FreeStroke(i,j)=FRX(i,j);
FreeStroke(i,j)=sqrt((FRX(i,j)+ShoulderWidth)^2+(FRY(i,j))^2)-OLength;
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%The force changes as the arm rotates from abduction to flexion
ForceNew(i,j)=(ForceDiff(i)/Rotation)*Beta2(i,j)+ForceAbd(i);


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%The actual stroke is the difference between the theoretical stroke
```

```
        %and the area of free stroke.
        ActualStroke(i,j)=TheoryStroke(i,j)- FreeStroke(i,j);
        %If the actual stroke is less then zero then the co-ordinates of
        %the actual stroke are also set to zero
        if ActualStroke(i,j)<0;
            ActualStroke(i,j)=0;
            GX2(i,j)=0;
            GY2(i,j)=0;
            GZ2(i,j)=0;
            %If the actual stroke is greater is greater then zero then the
            %co-ordinates have the free co-ordinates taken off to give the
            %resulting true value

        else
            GX2(i,j)=GX(i,j)- FGX(i,j);
            GY2(i,j)=GY(i,j)- FGY(i,j);
            GZ2(i,j)=GZ(i,j)-FGZ(i,j);
            FreeForceNew(i,j)=ForceNew(i,j);
        end
%           plot3(GX2(i,j),GZ2(i,j),GY2(i,j),':bs');
%           hold on
    end

    %ForceFle is the Flexion Force. it is decremented each iterations such
    %that it equals zero on the last iteration. This is the same for
    %Abduction as well
    ForceFle(i+1) = ForceFle(i)-DCountFle;
    ForceAbd(i+1) = ForceAbd(i)-DCountAbd;

    %Abduction and flexion increases for each iteration until it reaches
    %the maximum value
    Abduction2(i+1)=Abduction2(i)+AddA;
    Flexion2(i+1)=Flexion2(i)+AddF;
end


%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%The following calculations are necessary to find the mid point to compare
%values of the achievable shoulder outputs to the single value wrist
%outputs.
t=0
for Theta = 0:AddA:Abduction;
    t=t+1;

    %Abduction Capture counts the number of times up to when the stroke
    %separates from zero. Check the stroke length graph to understand.
    if ActualStroke(t,RotationCapture)<=0;
        AbductionCapture=AbductionCapture+1;
```

```
    end

end
i3=0
%The following section captures the average values of the stroke starting
%from the first non-zero value.
for i2 = AbductionCapture+1:i;
    i3=i3+1;
    AchievableStroke(i3) = ActualStroke(i2,RotationCapture);
    AchievableForce(i3)=FreeForceNew(i2,RotationCapture);
end
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

   Concept=('Concept 2')



   %Arm brace positions
    X5New(1)=X(1)
    Y5New(1)=X(2)

    X6New(1)=X(3)
    Y6New(1)=X(4)

    X7New(1)=X(5)
    Y7New(1)=X(6)

    X8New(1)=X(7)
    Y8New(1)=X(8)

    X9New(1)=0;
    Y9New(1)=-0.08;




%CALCULATION
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%Calculation for the spring length and Cable Length
    i=0;
    pop=0;
    pop1=0;
```

```
   pop2=0;
   pop3=0;
   R=0.5;

%Maximum Extension through to Maximum Flexion
for Theta = MaxE:-R:MaxF;
   i=i+1;
   Theta1(i) = Theta;

   %Hand Brace Positions
   Xnew2(i) = X2;
   Ynew2(i) = Y2;
%     Xnew4(i) = X4;
%     Ynew4(i) = Y4;


   %Calculates the position of the pivot respective to the hand and wrist
   Xnew1(i) = (X1*cos(Theta*3.16/180))-(Y1*sin(Theta*3.16/180));
   Ynew1(i) = (X1*sin(Theta*3.16/180))+(Y1*cos(Theta*3.16/180));
%     Xnew3(i) = (X3*cos(Theta*3.16/180))-(Y3*sin(Theta*3.16/180));
%     Ynew3(i) = (X3*sin(Theta*3.16/180))+(Y3*cos(Theta*3.16/180));

   HandMoment(i)=HandMass*Gravity*HandCOG*abs(cos(Theta*3.14/180));

BeamChange(i)=sqrt((Xnew1(i)-Xnew2(i))^2+(Ynew1(i)-Ynew2(i))^2)-sqrt((Xnew1(1)-
Xnew2(1))^2+(Ynew1(1)-Ynew2(1))^2);

if i==1
BeamChange2(i)=BeamChange(i);

else
   BeamChange2(i)=BeamChange(i)-BeamChange(i-1);
end



%Hand to cable
      Hipot5=(sqrt(X5New(1)^2+Y5New(1)^2));
      Gamma(i)=BeamChange2(i)/Hipot5;
      Psi5(i) = atan(Y5New(i)/X5New(i));
      Sigma5(i) = Gamma(i)+Psi5(i);
      Y5New(i+1) = Hipot5*sin(Sigma5(i));
      X5New(i+1) = Hipot5*cos(Sigma5(i));
      Ylong5(i) = Y5New(i)-(Xnew1(i)-Xnew2(i));
      Xlong5(i) = X5New(i);
      Delta5(i) = atan(Ylong5(i)/Xlong5(i));
      Beta5(i) = Psi5(i)-Delta5(i);
%spring
```

```
    Hipot6=(sqrt(X6New(1)^2+Y6New(1)^2));
    Psi6(i) = atan(Y6New(i)/X6New(i));
    Sigma6(i) = Gamma(i)+Psi6(i);
    Y6New(i+1) = Hipot6*sin(Sigma6(i));
    X6New(i+1) = Hipot6*cos(Sigma6(i));
    Ylong6(i) = (Y8New(1)-Y6New(i));
    Xlong6(i) = (X8New(1)-X6New(i));
    Delta6(i) = atan(Xlong6(i)/Ylong6(i));
    Beta6(i) = (pi()/2)-Delta6(i)-Psi6(i);
    Extend(i) = sqrt(Ylong6(i)^2+Xlong6(i)^2)-sqrt(Ylong6(1)^2+Xlong6(1)^2);

%input
    Hipot7=(sqrt(X7New(1)^2+Y7New(1)^2));
    Psi7(i) = atan(Y7New(i)/X7New(i));
    Sigma7(i) = Gamma(i)+Psi7(i);
    Y7New(i+1) = -Hipot7*sin(Sigma7(i));
    X7New(i+1) = -Hipot7*cos(Sigma7(i));
    Ylong7(i) = (Y9New(1)-Y7New(i));
    Xlong7(i) = (X9New(1)-X7New(i));
    Delta7(i) = atan(Ylong7(i)/Xlong7(i));
    Beta7(i) = Psi7(i)-Delta7(i);



    %Calculate the angle Omega for use to find the initial required
    %force to hold the hand in extension due to self weight. Omega is
    %the angle formed between the wrist and spring.
    %There are three sets of calculations necessary.
  SpringY(i) = Ynew2(i)-Ynew1(i); %Difference in the Y
  SpringX(i) = Xnew2(i)-Xnew1(i); %Difference in the X
    if Ynew1(i)>Ynew2(i)
      Gamma2(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
      Delta(i) = atan(abs(SpringY(i))/abs(SpringX(i)));
      Omega(i) = 3.14/2 - Gamma2(i) - Delta(i);
    elseif Ynew1(i)>0
      Gamma2(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
      Delta(i) = atan(abs(SpringX(i))/abs(SpringY(i)));
      Omega(i) = 3.14 - Gamma2(i) - Delta(i);
    else
      Gamma2(i) = atan(abs(Xnew1(i))/abs(Ynew1(i)));
      Delta(i) = atan(abs(SpringY(i))/abs(SpringX(i)));
      Omega(i) =  Gamma2(i) + Delta(i) - 3.14/2;
    end


  Force(i) = HandMoment(i)/((sqrt(Ynew1(i)^2+Xnew1(i)^2))*sin(Omega(i)));
  %Extension
```

```
    %The force in the elastic is a combination between the extension and
    %initial force.
    InstSpringForce(i)=(Force(i)*Hipot5*sin(Beta5(i))/(Hipot6*sin(Beta6(i))));
    SpringForce(i) = KSpring*Extend(i)+InstSpringForce(1); %Returns it to m from mm.


    RequiredStroke(i) = sqrt(Ylong7(1)^2+Xlong7(1)^2)-sqrt(Ylong7(i)^2+Xlong7(i)^2);


    RequiredForce(i)                                        =                                        ((-
Force(i)*Hipot5*sin(Beta5(i))+SpringForce(i)*Hipot6*sin(Beta6(i)))/(Hipot7*sin(Beta7(i))));



end



CaptureDiff=MaxE-MaxF;
CaptureDown=CaptureDiff/(i3-1);
i4=0;
for Theta = MaxE:-CaptureDown:MaxF;
    i4=i4+1;
    Theta4(i4) = Theta;
end


% %PLOT
% %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
% %Achievable Force versus Required Force
figure
%subplot(2,1,1)
hold on
plot(Theta1,RequiredForce)
plot(Theta4,AchievableForce,':r')
set(gca,'XDir','reverse')
title('Required (solid) and Achievable (hashed) Cable Force to hold wrist  statically')

ylabel('Cable Force (N)')
xlabel('Wrist Rotational Angle')
axis square
axis tight
grid on
hold off

%Achievable Stroke versus Required Stroke
%subplot(2,1,2)
figure
```

```
hold on
plot(Theta1,RequiredStroke)
plot(Theta4,AchievableStroke,':r')
set(gca,'XDir','reverse')
title('Required (solid) Achievable (hashed) Stroke Length in Meters (m)')

ylabel('Stroke Length (m)')
xlabel('Wrist Rotational Angle')
grid on
axis square
axis tight
hold off
%
%
% %Overall orthosis dimensions
% figure
% hold on
% title({'Overall Wrist orthosis Geometry';
%    [Concept];
%    ['X1 = ' ,num2str(X1),'  X2 = ' ,num2str(X2),'  X3 = ' ,num2str(X3),'  X4 = ' ,num2str(X4),];
%    ['Y1 = ' ,num2str(Y1),'  Y2 = ' ,num2str(Y2),'  Y3 = ' ,num2str(Y3),'  Y4 = ' ,num2str(Y4),];
%    ['Max Extension = ' ,num2str(MaxE),'  Max Flexion = ' ,num2str(MaxF),'  WristPiv1 = ' ,num2str(WristPiv1),'  WristPiv2 = ' ,num2str(WristPiv2),];
%    ['Spring Stiffness = ' ,num2str(KSpring),'  Mass = ' ,num2str(HandMass),'  COG = ' ,num2str(HandCOG),'  Efficiency = ',num2str(Efficiency),]})
% plot(X5New,Y5New)
% plot(X6New,Y6New,':b')
% plot(X7New,Y7New,'r')
% plot(X8New,Y8New,':rs')
% plot(CentreLineX, CentreLineY,'--mo')
% ylabel('Y Position (m)')
% xlabel('X Position (m)')
% axis square
% axis equal
% grid on
% hold off


BodyGeometriesX1 = [.035 WristPiv1 0.085 WristPiv2 -0.035]
BodyGeometriesY1= [-.2 0 .147 0 -.2]
LiftY = [Ynew2(1) Ynew1(1)]
LiftX = [(Y5New(1)+Xnew2(1)) Xnew1(1)]

BodyGeometriesX3 = [.035 WristPiv1 -0.085 WristPiv2 -0.035]
BodyGeometriesY3= [-.2 0 .147 0 -.2]
LiftY2 = [Ynew2(1) Ynew1(i)]
LiftX2 = [(Y5New(i)+Xnew2(1)) Xnew1(i)]
```

```
ArmGeometriesY = [-.2 0 .1 .1 0 -.2]
ArmGeometriesX = [.04 .03 .04 -.04 -.03 -0.04]

BeamX=[X5New(1) .0025 X7New(1) -.0025 X5New(1)]
BeamY=[Y5New(1) .0025 Y7New(1) -.0025 Y5New(1)]+Xnew2(1)
BeamX2=[X5New(i) -.0025 X7New(i) .0025 X5New(i)]
BeamY2=[Y5New(i) .0025 Y7New(i) -.0025 Y5New(i)]+Xnew2(1)

ShoulderCableX = [X7New(1) X9New]
ShoulderCableY = [Y7New(1) Y9New]+Xnew2(1)
ShoulderCableX2 = [X7New(i) X9New]
ShoulderCableY2 = [Y7New(i) Y9New]+Xnew2(1)

HandBeamX = [X5New(1) 0]
HandBeamY = [(Y5New(1)+Xnew2(1)) Xnew1(1)]
HandBeamX2 = [X5New(i) 0]
HandBeamY2 = [(Y5New(i)+Xnew2(1)) Xnew1(i)]

ElasticX = [X6New(1) X8New(1)]
ElasticY = [Y6New(1) Y8New(1)]+Xnew2(1)
ElasticX2 = [X6New(i) X8New(1)]
ElasticY2 = [Y6New(i) Y8New(1)]+Xnew2(1)

figure
subplot(2,2,1)
title('Plan View of Hand and Arm in Extension')
hold on
plot(Xnew2(1),0,'bx')
plot(BeamY,BeamX,'r')
plot(ShoulderCableY,ShoulderCableX,'m+:')
plot(ArmGeometriesY ,ArmGeometriesX,'r+:')
plot(ElasticY, ElasticX,'b+:')
plot(HandBeamY,HandBeamX,'b+-')
axis equal
set(gca,'YDir','reverse')
hold off


subplot(2,2,2)
hold on
title('Side View of Hand and Arm in Extension')
plot(BodyGeometriesY1,BodyGeometriesX1,'r+:')
plot(LiftX, LiftY,'b+-')
axis equal
hold off

subplot(2,2,3)
hold on
```

```
title('Plan View of Hand and Arm in Flexion')
plot(Xnew2(1),0,'bx')
plot(BeamY2,BeamX2,'r')
plot(ShoulderCableY2,ShoulderCableX2,'m+:')
plot(ArmGeometriesY,ArmGeometriesX,'r+:')
plot(ElasticY2, ElasticX2,'b+:')
plot(HandBeamY2,HandBeamX2,'b+-')
axis equal
set(gca,'YDir','reverse')
hold off

subplot(2,2,4)
hold on
title('Side View of Hand and Arm in Flexion')
plot(BodyGeometriesY3,BodyGeometriesX3,'r+:')
plot(LiftX2, LiftY2,'b+-')
axis equal
hold off
```

Appendix H

Alternative prototypes evaluated

1. Figure H.1 was the first concept created. It uses a linkage design which provides natural movement to the wrist. The design was considered impractical due to its unnatural appearance.

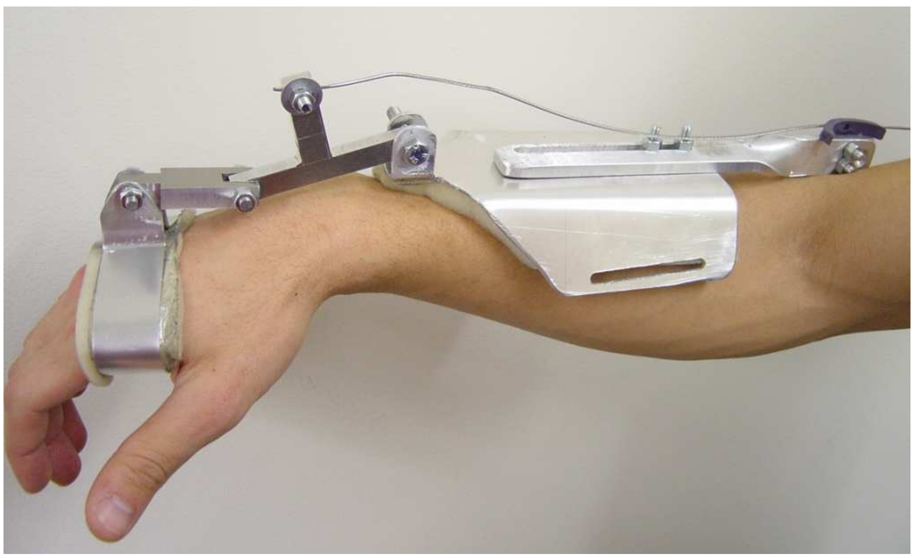


Figure H.1

2. Figure H.2 disguised the mechanism by placing it under the hand. However the rigid link design prevented the user from grasping many objects.

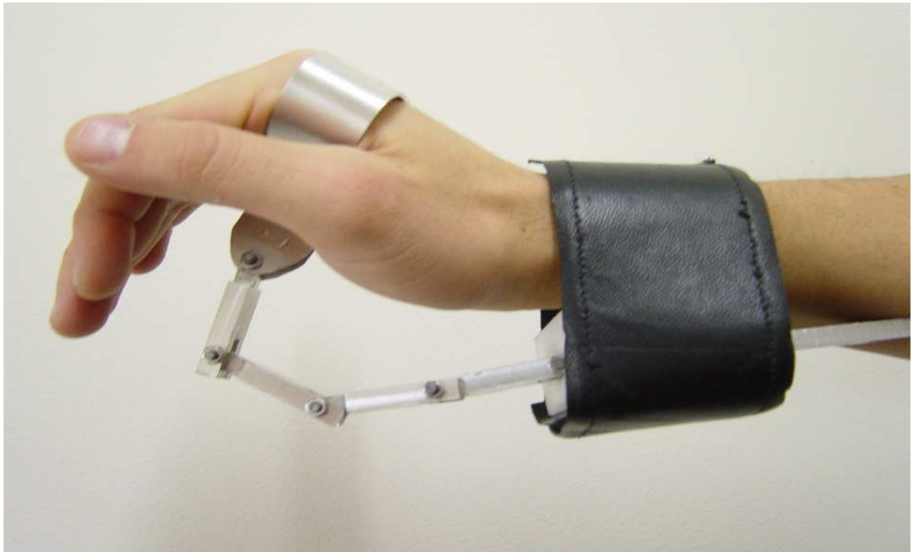


Figure H.2

3. Figure H.3 used a direct motor drive which moved the wrist easily. However this design required activation and controlled which was considered impractical in this research.

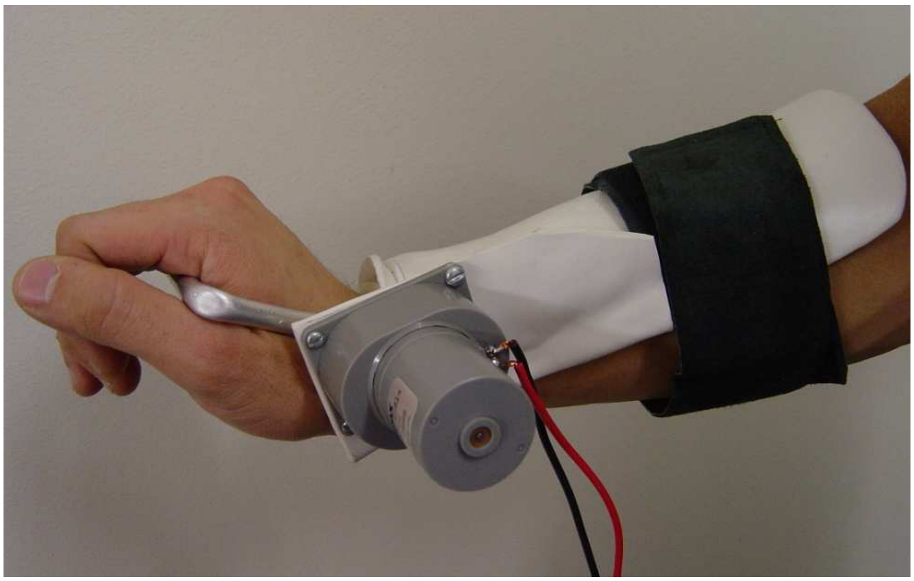


Figure H.3

4. Figure H.4 was the first system which attempted to use a spring which held the wrist in extension.

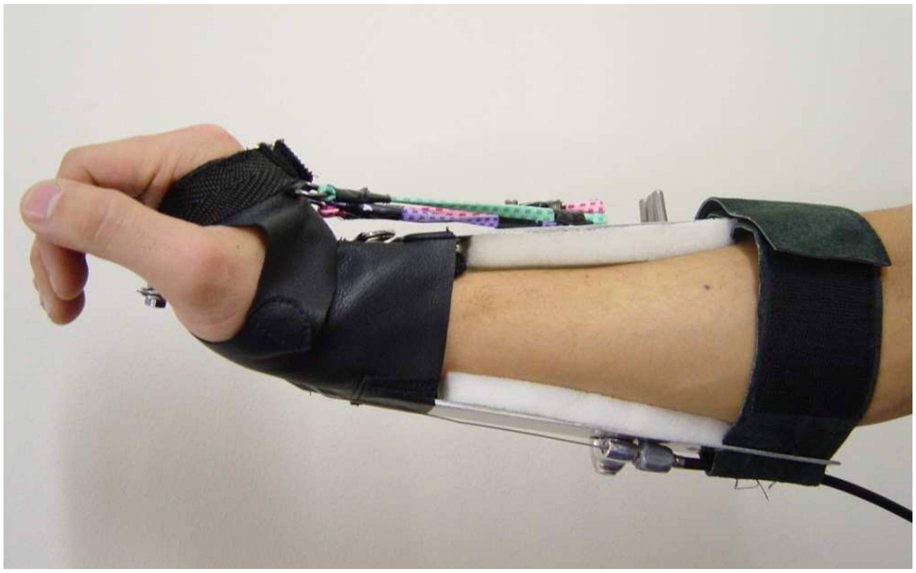


Figure H.4

5. The ‘Clock Spring’ design was used because it maintained a linear spring force. However the design was impractical due to the aesthetics.

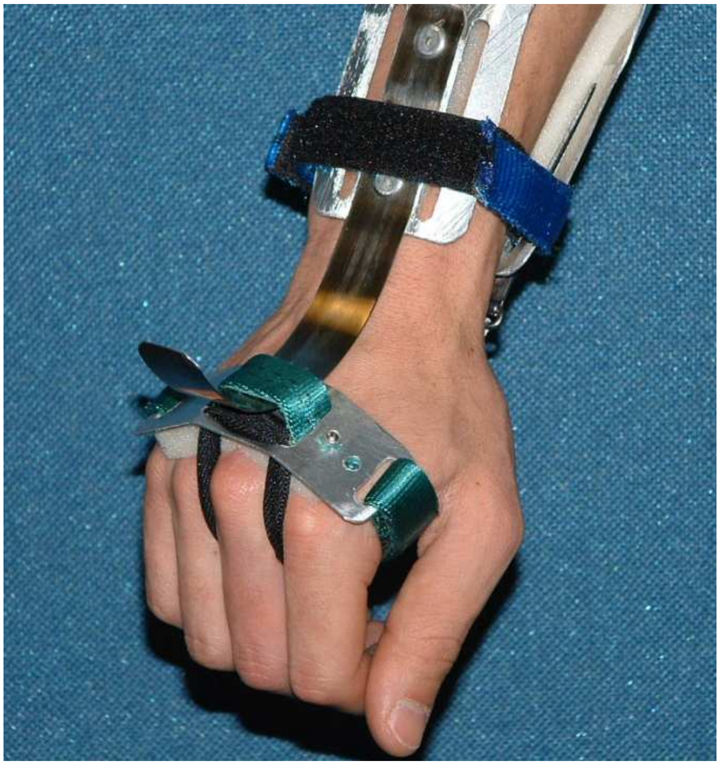


Figure H.5

6. Figure H.6 was functionally the same as the 'current concept' however the system was difficult to fit.

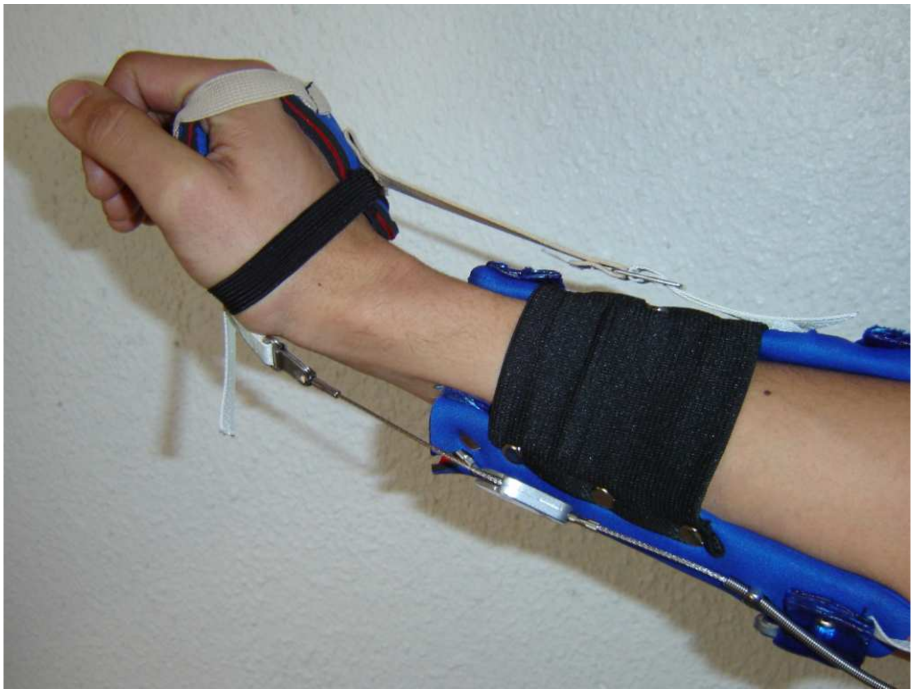


Figure H.6

7. Figure H.7 shows the final concept before the ‘current concept’ was created. The only problem with this system was the hand brace design which produced a large force between the index finger and the thumb.

Figure H.7